Experimental and numerical analysis of conventional and ultrasonically-assisted cutting of bone

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EXPERIMENTAL AND NUMERICAL ANALYSIS OF
CONVENTIONAL AND ULTRASONICALLY-
ASSISTED CUTTING OF BONE

by

Khurshid Alam

A Doctoral Thesis

Submitted in partial fulfilment of the requirements for the award of Doctor of
Philosophy of Loughborough University

Jan  2009

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Dedicated to My Mother (Mehrun Nisa)
Abstract

Bone cutting is widely used in orthopaedic, dental and neuro surgeries and is a technically demanding surgical procedure. Novel surgical methods are continually introduced in orthopaedic, neuro and dental surgeries and are aimed at minimising the invasiveness of the operation and allowing more precise cuts. One such method that utilises cutting with superimposed ultrasonic vibration is known as ultrasonically-assisted cutting (UAC). The main concern in bone cutting is the mechanical and thermal damage to the bone tissue induced by high-speed power tools. Recent technological improvements are concerned with the efforts to decrease the force required by the surgeon when cutting the bone as well as increases in surgery speed.

A programme of experiments was conducted to characterise properties of a bone and get a basic understanding of the mechanics of bone cutting. The experiments included: (a) nanonindentation and tension tests to obtain the properties for the finite element (FE) bone cutting model, (b) high-speed filming to observe the chip formation process, which influences thermomechanics of the cutting process in conventional drilling (CD) and ultrasonically-assisted drilling (UAD) and, (c) plane cutting and drilling experiments to measure the levels of force and temperature rise in the bone tissue.

Novel two-dimensional finite element (FE) models of cortical bone cutting were developed for conventional and ultrasonically-assisted modes with the MSC.MARC general FE code that provided thorough numerical analysis of thermomechanics of the cutting process. Mechanical properties such as the elastic modulus and strain-rate sensitivity of the bone material were determined experimentally and incorporated into the FE models. The influence of cutting parameters on the levels of stress, penetration force and temperature in the bone material was studied using conventional cutting (CC) and ultrasonically-assisted cutting (UAC). The temperature rise in the bone material near the cutting edge was calculated and the effect of cutting parameters on the level of thermal necrosis was analysed. The necrosis depth in bone was calculated as a distance from the cut surface to the point where the thermal threshold level was attained. Comparative studies were performed for the developed FE models of CC and UAC of bone and the results validated by conducting experiments and using data from scientific publications.
The main outcome of the thesis is an in-depth understanding of the bone cutting process, and of its possible application in orthopaedics. Recommendations on further research developments are also suggested.

**KEYWORDS:** Bone cutting, Cortical bone, Ultrasonic vibration, Thermomechanics, Experimental methods, Finite element analysis, Thermal Necrosis
Acknowledgements

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I owe many thanks to my Co-supervisor, Prof. Vadim V. Silberschmidt, for allowing me the freedom to pursue my own directions in this research throughout my time at Loughborough. His expert guidance, mentorship, insight and experience undoubtedly helped in the fulfilment/accomplishment of this undertaking.

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I acknowledge with much appreciation the role of academic, technical and administration staff of the Wolfson School of Mechanical and Manufacturing Engineering. And a deep-thank to Dr. Allan Meadows for his advice, useful ideas and practical help in all of experiments performed on ultrasonic cutting system.

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I would like to appreciate the team of “Stephen Morris” the meat shop in Loughborough town centre, for generous and prompt supply of fresh and cleaned bovine bone specimens.
I wish to express my love and gratitude to all my family members, friends for their endless love through the duration of my studies.
## Notation and abbreviations

### Symbols

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Meaning</th>
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<tbody>
<tr>
<td>$A$</td>
<td>Amplitude of vibration, $\mu$m</td>
</tr>
<tr>
<td>$A_o$</td>
<td>Area faction of osteonal bone</td>
</tr>
<tr>
<td>$A_m$</td>
<td>Area faction of interstitial matrix</td>
</tr>
<tr>
<td>$C_b$</td>
<td>Heat capacity of bone, J/m$^3$K</td>
</tr>
<tr>
<td>$D$</td>
<td>Drill diameter, mm</td>
</tr>
<tr>
<td>$E$</td>
<td>Young’s modulus, GPa</td>
</tr>
<tr>
<td>$E_b$</td>
<td>Young’s modulus of bone, GPa</td>
</tr>
<tr>
<td>$E_o$</td>
<td>Young’s modulus of osteon, GPa</td>
</tr>
<tr>
<td>$E_m$</td>
<td>Young’s modulus of interstitial matrix, GPa</td>
</tr>
<tr>
<td>$E_i$</td>
<td>Young’s modulus of indenter, GPa</td>
</tr>
<tr>
<td>$E_r$</td>
<td>Reduced Young’s modulus, GPa</td>
</tr>
<tr>
<td>$F_c$</td>
<td>Force along the tool face</td>
</tr>
<tr>
<td>$F_s$</td>
<td>Force along the shear plane</td>
</tr>
<tr>
<td>$F_o$</td>
<td>Force in osteon</td>
</tr>
<tr>
<td>$F_m$</td>
<td>Force in interstitial matrix</td>
</tr>
<tr>
<td>$F_Q$</td>
<td>Vertical component of force</td>
</tr>
<tr>
<td>$F_p$</td>
<td>Horizontal component of force</td>
</tr>
<tr>
<td>$f$</td>
<td>Frequency of vibration, kHz</td>
</tr>
<tr>
<td>$f_r$</td>
<td>Feed rate, mm/min</td>
</tr>
<tr>
<td>$G$</td>
<td>Shear modulus, GPa</td>
</tr>
<tr>
<td>$G_{cs}$</td>
<td>Toughness, J/m$^2$</td>
</tr>
<tr>
<td>$h$</td>
<td>Convective heat transfer coefficient, W/m$^2$K</td>
</tr>
<tr>
<td>$H$</td>
<td>Contact heat conduction coefficient, W/m$^2$K</td>
</tr>
<tr>
<td>$K$</td>
<td>Shear yield stress, MPa</td>
</tr>
<tr>
<td>$K_b$</td>
<td>Thermal conductivity of bone, W/m K</td>
</tr>
<tr>
<td>$K_{tc}$</td>
<td>Fracture toughness, MN/m$^{3/2}$</td>
</tr>
<tr>
<td>$L_c$</td>
<td>Length of tool-chip contact</td>
</tr>
<tr>
<td>$N$</td>
<td>Drilling speed, rpm</td>
</tr>
<tr>
<td>$N_c$</td>
<td>Force along the tool face</td>
</tr>
<tr>
<td>$N_s$</td>
<td>Force perpendicular to the tool face</td>
</tr>
<tr>
<td>$R$</td>
<td>Reaction for between the chip and tool</td>
</tr>
</tbody>
</table>
$R_a$ Average roughness, µm
$r$ Contact ratio
$t_a$ Start time of tool work piece contact in vibration cutting
$t_b$ End time of tool work piece contact in vibration cutting
$T$ Temperature, ºC
$T_{melt}$ Melting temperature, ºC
$T_i$ Cutting tool temperature, ºC
$T_b$ Bone temperature, ºC
$T_a$ Ambient temperature, ºC
$t_1$ Uncut chip thickness, mm
$t_2$ Chip thickness, mm
$V_c$ Cutting speed, mm/sec
$\Delta T$ Temperature rise, ºC
$\nu_{cr}$ Critical sliding velocity, mm/s
$\nu_r$ Relative sliding velocity, mm/s
$\alpha$ Rake angle, °
$\beta$ Friction angle, °
$\psi$ Tool nose radius, micrometers
$\dot{\varepsilon}$ Strain rate, s$^{-1}$
$\bar{\varepsilon}_p$ Equivalent plastic strain
$\varepsilon_p$ Plastic strain
$\dot{\varepsilon}_o$ Reference strain rate
$\varepsilon_m$ Strain in interstitial matrix
$\varepsilon_o$ Strain in osteon
$\dot{\varepsilon}_p$ Plastic strain rate, s$^{-1}$
$\phi$ Shear angle, °
$\mu$ Coefficient of friction
$\bar{\mu}$ Mean coefficient of friction
$\sigma_{ys}$ Compressive strength, MPa
$\sigma_{yt}$ Tensile yield strength, MPa
$\sigma_o$ Stress in osteon, MPa
$\sigma_m$ Stress in interstitial matrix, MPa
$\bar{\sigma}$ Equivalent stress, MPa
$\nu$ Poisson’s ratio
$\nu_{bone}$ Poisson’s ratio of the bone
$\rho$ Density, kg/m$^3$
### Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Meaning</th>
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<tbody>
<tr>
<td>ALE</td>
<td>Arbitrary Lagrangian Eulerian</td>
</tr>
<tr>
<td>CC</td>
<td>Conventional cutting</td>
</tr>
<tr>
<td>CD</td>
<td>Conventional drilling</td>
</tr>
<tr>
<td>CPC</td>
<td>Conventional plane cutting</td>
</tr>
<tr>
<td>DOC</td>
<td>Depth of cut</td>
</tr>
<tr>
<td>DTN</td>
<td>Depth of thermal necrosis</td>
</tr>
<tr>
<td>FE</td>
<td>Finite element</td>
</tr>
<tr>
<td>FEA</td>
<td>Finite element analysis</td>
</tr>
<tr>
<td>FEM</td>
<td>Finite element method</td>
</tr>
<tr>
<td>JC</td>
<td>Johnson-Cook</td>
</tr>
<tr>
<td>NTL</td>
<td>Necrosis threshold level</td>
</tr>
<tr>
<td>SEM</td>
<td>Scanning electron microscope</td>
</tr>
<tr>
<td>UAC</td>
<td>Ultrasonically assisted cutting</td>
</tr>
<tr>
<td>UAPC</td>
<td>Ultrasonically assisted plane cutting</td>
</tr>
<tr>
<td>UAD</td>
<td>Ultrasonically assisted drilling</td>
</tr>
</tbody>
</table>
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Conference Contributions


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Chapter 1. Introduction

1.1 Project Background

Bone fracture is a feature of everyday life due to accident or aging. When a bone is broken the periosteum (outer surface) and endosteum (inner surface touching the marrow) provide bone-forming cells, which endeavour to bridge the fracture. The fracture of the bone is usually covered by drilling the bone at required sites for screw insertion and fixing plates.

The cutting of bone is one of the oldest surgical procedures in the history of medicine. Trepanation, an operation to remove a portion of the skull bone, may have been performed as far back as 10,000 B.C. [1]. Knee and hip implant surgeries are performed around the world and are considered the most common operations in clinical practice. A total of 300,000 knee arthroplasties are performed each year in the United States alone with the number increasing every year [2]. Joint replacement and fixation of the parts to perform kinematic and dynamic function of a human joint is a known treatment in bone surgery. Also bone machining for manufacturing of bone screws to replace metal ones is becoming popular. It is expected that bone screws will be useful for bone healing that will enable micro movements at fracture sites and will minimise per-implant osteopenia (a condition where bone’s mineral density is lower than normal) [3]. The different methods of bone cutting include scraping, grooving, sawing, drilling, boring, grafting and shearing. Among these methods, drilling is a surgical operation, most discussed in literature.

Bone cutting has always been a problem for surgeons due to its sensitivity to many osteotomes. A considerable manual force is required by a surgeon to cut the bone either with a chisel or producing a hole for fixation purposes with a rotational tool such as a drill. Cutting of bone with conventional tools such as oscillating saws and chisels require large exposure of the donor site with associated trauma to nerves and muscles [4]. Harvesting long bone grafts may also cause pain due to stress fractures in the bone around the donor site [5]. Another concern in such incisions is the heat generation in a high speed cutting process, which may induce thermal damage and weakening of the fixative devices anchoring the bone. The high speed cutting process such as drilling, produces heat and can result in trauma. It can also cut inefficiently because the cutting flutes become clogged. Temperature and force measurements are important in predicting the thermal necrosis threshold level and for control of the cutting tool to avoid unnecessary damage to the soft tissues.
A bone-tool interaction in conventional cutting (CC) and ultrasonically-assisted cutting (UAC) modes has been of interest to researchers for the last few decades. The advantages and disadvantages of both modes have been reported in literature, making selection of a cutting tool (conventional or ultrasonic) for orthopaedics, dentistry and neurosurgery difficult. Presently, a mechanical rotary drill is the main type of drilling equipment used in clinical practices. Various drilling techniques have been introduced to improve the process in order to minimise the invasiveness of the operation. One such modern drilling technique utilises high-frequency, ultrasonic vibration of the drill along its longitudinal axis and is called ultrasonically-assisted drilling (UAD). Ultrasonic vibration has been already successfully applied on a wide scale in cutting high-strength aerospace alloys [6], composites [7] and soft materials [8]. Ultrasonically-assisted surgical instruments for cutting bones and removing the periosteum are becoming popular in orthopaedic, neuro and dental surgeries [1]. In medical applications an ultrasonic tool can reduce cutting forces and provide a surgeon with better control to cut the bone tissue [9]. This technique allows significant improvements in processing intractable materials with a multifold decrease in cutting forces; an improvement in surface finish can be achieved by the use of ultrasonic cutting.

Investigation of cutting procedures and parameters is necessary to evaluate the damage to the bone and to improve bone-implant/screw interfacial strength. Modeling bone cutting using finite-element analysis (FEA) may be considered as a promising and reliable technique for the development of new surgical procedures in the near future. FE method has been utilised to study the response of bone at macro and micro levels under various loading conditions, but is not yet adequately applied to model bone cutting. In fact, finite element studies describing thermo-mechanics of bone cutting have not been reported in the literature to date.

A detailed analysis of the bone cutting process necessitates a combination of experimental studies and numerical simulations of the cutting tool – bone interaction. In this research work, all the shortcomings of the previous studies were identified from a comprehensive literature survey and a series of experiments and numerical simulations were conducted to analyse the structural and thermal processes involved in bone cutting. Particular attention was paid to the benefits of UAC and the nature of the obtained improvements in cutting of bone compared to CC. Since the ultrasonically-assisted technique was previously proposed as an enhancement of the existing CC methods, the comparison is performed with CC at some stage of the research.
1.2 Aim and Objectives

Aims:

The main aim of this research study was to get in-depth knowledge of the mechanics of orthopaedic tool penetration into cortical bone tissue with conventional and ultrasonically-assisted modes and to provide engineering-based information in order to improve current surgical procedures. It is hoped that the results of this research will be useful for optimisation of the bone cutting procedure and extension of the use of the ultrasonically supported technology in orthopaedic surgeries.

It is worth pointing out that the experiments carried out in this thesis are mainly focused on the study of the basic understanding of the bone cutting process, and mechanical tests provide necessary data for the development of a FE bone cutting model. Most of the experiments described here are unique (to the author’s knowledge) and aimed at gaining advanced understanding of the UAC of bone.

Finite element (FE) modelling can provide detailed information on the mechanical and thermal features of the cutting process. To the author’s knowledge, no FE models so far have been suggested for bone cutting. Hence, one of the objectives of the research was to develop a FE model for two cutting techniques – CC and UAC, as well as to investigate the influence of various cutting parameters on thermomechanics of bone cutting.

Objectives:

The objectives of the research are:

1. to study bone cutting mechanics for linear and rotational tools;
2. to investigate the influence of the tool’s ultrasonic vibrations on the cutting process; and possible applications of ultrasonic tools in orthopaedics;
3. to measure experimentally penetration forces in conventional and ultrasonic bone cutting;
4. to measure bone cutting (drilling) temperatures experimentally using thermocouples;
5. to acquire cortical bone properties for the FE bone cutting model;
6. to develop a fully thermomechanically coupled FE cutting model capable of simulating the kinematics of conventional and vibrated tools and to predict the level of cutting forces, stresses and temperature in bone tissue;
7. to investigate, both experimentally and using FE simulations, the influence of various cutting parameters on the thermomechanics of bone cutting;
8. to compare obtained FE and experimental results.

All the results for the features and parameters of bone cutting, obtained from experimental work and the FE study, will be compared with those available in the literature.

1.3 Outline of the Report

The present work incorporates 9 other chapters, the summary of which is the following:

**Chapter 2** deals with the mechanical and thermal properties of bone tissue, bone microstructure and inelastic deformation in cortical bone.

**Chapter 3** describes the basics of the cutting process and explains various bone cutting methods. The chapter also reviews the methods used so far in investigating the optimum cutting parameters in bone cutting. Orthopaedic tools and materials are also discussed.

**Chapter 4** presents a general theory of FEM followed by the literature review of the development of FE cutting models. Some recent studies on modelling UAC are also introduced.

**Chapter 5** describes experimental methods used in acquiring bone material properties. These includes tension tests and nanoindentation.

**Chapter 6** focuses on the measurements of forces produced in plane cutting and drilling with and without ultrasonic vibrations. The effect of a vibrated drill on the chip formation mechanism, chip shape and drilled hole surface quality is briefly discussed.

**Chapter 7** is concerned with the measurement of drilling temperatures using thermocouples.

**Chapter 8** gives details of the general features and development of simulation models for both CC and UAC. Modelling of the bone material, boundary conditions and cooling environment (irrigation) are presented.

**Chapter 9** presents a detailed discussion of the results obtained with the developed FEM model. The discussion is focused on the effect of various cutting parameters such as friction and cutting speed on the stress distributions, cutting force and temperature as well as the effects of the ultrasonic frequency and amplitude on the cutting force. The chapter also compares the results of bone cutting experiments and those obtained in numerical simulations.
Chapter 10 summarizes the outcomes of the research, presents conclusions drawn on the basis of the experimental results and numerical simulations, and recommends possible further work on the topic.

Note: In the subsequent chapters 2, 3 and 4 of the literature review, the type of bone is described whenever possible.
Chapter 2. Cortical Bone Tissue and Its Properties

2.1 Introduction

The mechanical characteristics of cortical bone, especially those related to age and diseases, are of large interest to the biomedical community. The overall mechanical properties of bone is the contribution of each level of structural elements organized in a hierarchical fashion [10][11]. From the perspective of materials science, an advanced understanding mechanics of bone is necessary for the development of biocompatible bone substitute with load bearing capacity. Such substitute will be helpful in developing more accurate numerical and analytical models for the analysis of bone-replacement materials.

Bone has a complex microstructure compared to other engineering materials and carries unique properties such as self-repair and adaptation to changes in external mechanical loadings [12]. A great amount of information has already been accumulated on the mechanical behaviour of bone and its stiffness, strength and toughness have been shown to be affected by its components at the architectural level and molecular moments. The exact properties of bone material particularly thermal and fractural properties are necessary for future analytical and FE models of bone cutting.

2.2 Bone Tissue

Bones are rigid organs composed of an organic protein and the inorganic mineral hydroxyapatite. The natural shapes of the bones and its unique shape and structure, enables them to be lightweight and strong, while performing other functions involving growth. These properties are attributed to a complex combination of constituents forming a bone. Bone exhibits an elastic behaviour with a lower level of stiffness than that of metals, is stronger than engineering plastics and with better ductility than most ceramics [13]. The functions of bone can be categorised into two types – physiological and mechanical. Physiologically, bone provides storage for calcium, salts and minerals for the body and produces blood cells in the marrow. Mechanically bone provides support and protection to the body and its organs. Ligaments and tendons are often attached to bones enabling them to act as a lever system to provide movement and locomotion. After more than a century of study, scientists are still only beginning to get understanding of bone’s features.

Based on the shape, bones are categorised into three primary groups: long, short or flat bones. Femur and tibia, which consist of a hollow tubular region containing marrow, are
examples of long bones. Short bones such as carpals and vertebral bodies are of irregular shape, are associated with more complex movements and have a relatively thin cortical layer. Flat bones have entirely different dimensions and structural functions from the other two. Examples of flat bones are skull, scapula and mandible. Long bones, such as tibia or femur can be divided into two distinct parts: the epiphyses and the diaphysis (see Figure 2.1a). The diaphysis is linked to epiphysis and consists largely of cortical bone. Epiphysis is the end region of long bones composed of both cortical and spongy bone allowing growth with the least possible loss of strength.

**Cortical and trabecular bone**

Two main types of bone tissue are cortical (compact) bone and trabecular (spongy) bone. Cortical bone is more compact, load-bearing and impact-resistant. Cortical bone constitutes elongates such as the femur or tibia those act as main load-bearing members of the skeleton. Cortical bone such as the cranium or ribs also protects the softer inner organs. Trabecular bone is a much more porous tissue with cellular solids, and exhibits lower effective strength than

![Figure 2.1](a) Femur bone (b) cortical and trabecular bone

than cortical bone. In studies of the measurements of the elastic modulus and compressive strengths of these two types of bone, cortical bone has been found to be stronger than trabecular bone. Because of this, trabecular bone has to be thicker in comparison in order to meet the demands placed upon it, such as withstanding compressive stresses or dampening of
loads through the surfaces of contacting joints [14]. The anatomical divisions of a long bone and locations of cortical and trabecular bone are shown in Figure 2.1. Trabecular bone is of complex microstructure with porosity ranging from 50% to 90%. Since finite element modeling of trabecular bone is much more complex than that of cortical bone (which is less porous), therefore, this research was mainly focused on cortical bone as a starting point.

2.3 Bone Microstructure

Bone is a material with different hierarchical levels with various structural components. Components of lower levels serve as the building blocks for higher levels. A detailed classification of the structural hierarchy across the varying length scales, has been proposed independently by Weiner [10] and Rob [11] (Figure 2.2). These levels and structures are: (1) the macrostructure: cancellous and cortical bone; (2) the microstructure (from 10 to 500 µm): Haversian systems, osteons, single trabeculae; (3) the sub-microstructure (1–10 µm): lamellae; (4) the nanostructure (from a few hundred nanometers to 1µm): fibrillar collagen and embedded minerals and (5) the subnanostructure (below a few hundred nanometers): a molecular structure of constituent elements, such as mineral, collagen, and non-collagenous organic proteins. The hierarchically organized structure and the placement of the components of bones, make them heterogeneous and anisotropic [11].
Classification of whole bones depends mainly on the anatomical location (e.g., femur, tibia, etc.). Up to three main classes of cortical bone are defined as lamellar, osteonal and woven bone. Lamellar bone is a laminated structure similar to plywood [15]. Osteonal bone, or Haversian bone, has cylindrical lamellae surrounding a Haversian canal. Haversian and Volkmann's canals are interconnected in the cortical bone tissue and serve the purposes of the cardiovascular and nervous systems. Osteons may be considered as fibre reinforcements in the composite structure of the bone and are generally aligned in its principal direction. The remainder of the cortical bone is called interstitial bone. A more disordered woven bone, which is found in young individuals and has different mechanical properties than lamellar bone. With age, woven bone transforms this tissue into a mature osteonal bone by the process of remodelling [15].

2.4 Bone as a Natural Composite Material

Bone is generally considered as a composite material consisting of an elastic mineral ‘fibres’ embedded into an organic matrix having pores filled with liquids. Cortical and trabecular bone are commonly reported to have either transversely isotropic or orthotropic material properties. Lamellar bone tissue can be considered as a fibre-reinforced composite of collagen fibrils and mineral crystals. At the microstructural level, a Haversian bone tissue can be considered as a set of secondary osteons embedded in the matrix of interstitial lamellae. The osteonal bone is separated from the matrix by a thin interface (cement line). Several authors have modelled bone as a standard fibre – rod – or platelet-reinforced material and applied theories of composite materials to bone tissue with various contents of phases [16][17][18]. Pidaparti [17] suggested that osteonal bone can be modelled as simple reinforced composite, provided that accurate properties for the mineral and collagen phases of the ultrastructure are available.

2.5 Mechanical/Structural Properties of Cortical Bone

The quantification of mechanical properties of bone can be useful in the development of theories concerning its fracture mechanisms. One of the important characteristics of cortical bone is its apparent density ranging from 1.7 and 2.0 g cm$^{-3}$ and its porosity varying from 30% to 90% [19]. Similar to other engineering materials, bone responds differently to quasistatic and dynamic loadings. Damage and failure of bovine compact bone have already
been studied in quasi-static tensile experiments and have shown large variations of stress failure, from 100 to 200 MPa, and failure strain, from 0.4% to 4% [20][21][22].

The elastic modulus of cortical bone tissue has been shown to increase with an increase in loading rate [23][24]. Bone shows an asymmetric behaviour with regard to the longitudinal resistance, being in compression much higher than in tension. The mechanisms by which bone failure occurs are different from isotropic materials due to its hierarchical structure and composition. The study of continuum and multi-scale models that are able to capture all the structural and thermal features of bone, will be useful in the development of advanced models of bone cutting. Such studies will also be helpful in the development of alternative biomimetic materials. The characterisation of mechanical properties of bone is essential and will be a vital input to the analytical and numerical analysis of bone cutting.

2.5.1 Anisotropy of bone

Morphology of the cortical bone is well reported in the literature and the overall elastic moduli have been measured using various types of experiments. In orthotropic material, properties are independent of direction within two mutually perpendicular plane of symmetry. Such materials are defined by nine independent variables (i.e. elastic constants). On the other hand, transverse isotropic materials are those that have similar properties in one plane and are different in the direction perpendicular to that plane. Such materials are defined by five independent elastic constants. Most of the recent studies have discussed bone as an orthotropic material [25][26]. The constitutive equation for an orthotropic elastic material is the conventional form of the Hooke's Law,

$$\sigma_i = C_{ij} \varepsilon_j,$$ \hspace{1cm} (2.1)

Where $\sigma_i$ is the stress tensor in one-index notation, $\varepsilon_j$ is the strain tensor and $C_{ij}$ is the stiffness tensor, which for the orthotropic case is positive definite and symmetric:

$$C_{ij} = \begin{bmatrix}
C_{11} & C_{12} & C_{13} & 0 & 0 & 0 \\
C_{12} & C_{22} & C_{23} & 0 & 0 & 0 \\
C_{13} & C_{23} & C_{33} & 0 & 0 & 0 \\
0 & 0 & 0 & C_{44} & 0 & 0 \\
0 & 0 & 0 & 0 & C_{55} & 0 \\
0 & 0 & 0 & 0 & 0 & C_{66}
\end{bmatrix}.$$
Figure 2.3 represents the bone’s coordinate system where $X_1$ is the radial bone direction, $X_2$ is in the circumferential bone direction and $X_3$ is the longitudinal bone direction. Transverse isotropy assumes $X_3$ to be the axis of symmetry. This assumption also modifies the stiffness matrix, reducing the number of independent coefficients from nine to five. Stress-strain relationships can also be expressed in terms of the compliance tensor that is the inverse of the stiffness tensor. For an orthotropic material, the compliance and stiffness tensors are related to the material properties of Young's modulus $E_i$, the Poisson's ratio $\nu_{ij}$ and the shear modulus $G_{ij}$ as:

\[
S_{ij} = C_{ij}^{-1} = \begin{bmatrix}
1/E_i & -\nu_{21}/E_2 & -\nu_{31}/E_3 & 0 & 0 & 0 \\
-\nu_{12}/E_1 & 1/E_2 & -\nu_{32}/E_3 & 0 & 0 & 0 \\
-\nu_{13}/E_1 & -\nu_{23}/E_2 & 1/E_3 & 0 & 0 & 0 \\
0 & 0 & 0 & 1/G_{23} & 0 & 0 \\
0 & 0 & 0 & 0 & 1/G_{31} & 0 \\
0 & 0 & 0 & 0 & 0 & 1/G_{12}
\end{bmatrix}.
\]

The poisson’s ratio $\nu_{ij}$ is expressed as $-\frac{\varepsilon_i}{\varepsilon_j}$ for a strain in the $i$ direction, and $G_{ij}$ are the shear moduli in the $i,j$ planes. The symmetry condition requires

\[
\frac{\nu_{ij}}{E_i} = \frac{\nu_{ji}}{E_j}.
\]  

(2.2)

In a recent study by Öriás [15], the elastic constants of human femur in three orthogonal directions with longitudinal waves, were measured using ultrasonic characterisation The
results showed that bone exhibits overall orthotropy as its elastic symmetry, but the midshaft location results suggest local transverse isotropy.

2.5.2 Elastic properties of bone

There are a number of studies devoted to the measurements of the elastic properties of bones. Most of the studies used experimental techniques to measure these properties. A review of the cortical bone’s anisotropy and measurements techniques for elastic constants reported in literature are given in Table 2.1 together with magnitudes of these constants.

In research by Taylor [28], the CT data of cadaveric human bone and its natural frequencies were obtained using modal analysis. From the geometry of the CT data, a finite element model was produced with the distribution of density established by calculating the mass of the FE bone model and comparing it with the mass of the bone. The values of the orthotropic elastic constants were then established by matching the calculations of FE analyses with those obtained with natural frequencies used in experiments, predicting a maximum error of 7.8% over 4 modes of vibration.
Table 2.1 Elastic properties and anisotropy of bone. $E$ – MPa, $G$ – MPa

<table>
<thead>
<tr>
<th>Reference</th>
<th>Bone Type</th>
<th>Test Method</th>
<th>$E_1$</th>
<th>$E_2$</th>
<th>$E_3$</th>
<th>$G_{12}$</th>
<th>$G_{13}$</th>
<th>$G_{23}$</th>
<th>$v_{12}$</th>
<th>$v_{13}$</th>
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<th>$v_{21}$</th>
<th>$v_{31}$</th>
<th>$v_{32}$</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>[25]</td>
<td>Bovine</td>
<td>Ultrasonic</td>
<td>20</td>
<td>26.2</td>
<td>3.14</td>
<td>4.5</td>
<td>-</td>
<td>0.2</td>
<td>0.22</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>Transversely isotropic</td>
</tr>
<tr>
<td>[26]</td>
<td>Human</td>
<td>Ultrasonic</td>
<td>12.6</td>
<td>12.9</td>
<td>21.2</td>
<td>4.4</td>
<td>5.5</td>
<td>6.1</td>
<td>0.419</td>
<td>0.24</td>
<td>0.23</td>
<td>0.43</td>
<td>0.4</td>
<td>0.38</td>
<td>Orthotropic</td>
</tr>
<tr>
<td>[27]</td>
<td>Bovine</td>
<td>Ultrasonic</td>
<td>18.7</td>
<td>20</td>
<td>28</td>
<td>2.9</td>
<td>2.8</td>
<td>3.7</td>
<td>0.26</td>
<td>0.17</td>
<td>0.17</td>
<td>0.28</td>
<td>0.26</td>
<td>0.25</td>
<td>Orthotropic</td>
</tr>
<tr>
<td>[28]</td>
<td>Human</td>
<td>Ultrasonic</td>
<td>17.9</td>
<td>18.1</td>
<td>22.8</td>
<td>7.11</td>
<td>6.58</td>
<td>5.71</td>
<td>0.26</td>
<td>0.3</td>
<td>0.31</td>
<td>0.28</td>
<td>0.38</td>
<td>0.37</td>
<td>Orthotropic</td>
</tr>
<tr>
<td>FEA</td>
<td></td>
<td></td>
<td>13.4</td>
<td>14.1</td>
<td>22.9</td>
<td>6.20</td>
<td>5.8</td>
<td>4.6</td>
<td>0.42</td>
<td>0.23</td>
<td>0.23</td>
<td>0.42</td>
<td>0.23</td>
<td>0.23</td>
<td>Orthotropic</td>
</tr>
</tbody>
</table>
2.5.3 Inelastic behaviour of cortical bone

The characterisation of the specific mechanical behaviour of bone tissue entails the formulation of a representative constitutive model. The mechanical properties of cortical bone have been extensively studied at the macrostructural scale. The study of inelastic behaviour of bone is essential as this phenomena occurs between prosthetic devices and bone, such as in orthopaedic or oral implantology [29].

A number of studies of both cortical and trabecular bone have demonstrated their inelastic strain [20][29][30]. Mercer [30] investigated a mechanical behaviour of bovine femur cortical bone relating the load to the axial strain in monotonic testing to failure using flexural test. All tests provided approximately the same inelastic strains and strain hardening. The Young’s modulus (E = 25 GPa) was the same in tension and compression, and consistent with reported magnitudes. The yield strains in tension (0.6%) and compression (1.1%) were also consistent with those reported. Experiments identified dilatation as well as tension/compression asymmetry and the Drucker-Prager constitutive law was proposed for cortical bone’s behaviour. In tension, the material behaved as an inelastic solid with linear strain hardening (Figure 2.4: tangent modulus, $E_H = 1$ GPa). In compression, the peak stress was appreciably larger than in tension, but the strain at failure was smaller in compression (1.6% in compression and 2.4% in tension). Similar behaviour of cortical bone was observed in [20] (see Figure 2.5a). In tension, it yielded, followed by linear hardening up to a failure strain of approx. 2.8%. In compression, the yield strength was higher than that in tension and after the peak the tissue softens up to the measured strain of 1.8% at failure. The inelastic strain has been attributed to the development of diffuse microcrack arrays [31][32]. In compression, the softening has been linked to the formation of shear bands [32].

**Figure 2.4** Tensile and compressive stress/strain curves for cortical bone ascertained from flexure test [30]
DePaula [21] tested bovine femur cortical bone specimens in tension after being treated in-vitro for 3 days with sodium fluoride solutions of different molarity (0.145, 0.5 and 2.0 M). The tensile testing showed a decreased elastic modulus, yield stress, ultimate stresses and increased ultimate strains and toughness with increased fluoride treatment concentration as shown in Figure 2.5(b). Typical stress-strain graph obtained from each test group showed this behaviour.

Figure 2.5 Inelastic response of cortical bone: (a) schematic of tensile and compressive stress/strain curves for cortical bone along longitudinal axis [20]; (b) typical stress–strain curves from four groups of bovine femur cortical bone treated with NaF: control, 0.145 M, 0.5 M, and 2.0 M [21]

A micromechanical model for cortical bone tissue using an overlapped platelets model was produced by Kotha [22]. The organic matrix was considered to be elastic in tension and elastic-plastic in shear until it failed. The model suggested that the elastic modulus and
shear yield stress of the organic matrix was correlated with the bone mineral content (BMC), up to the point when it yielded. The model implied that reinforcement of the organic matrix by mineral platelets affect pre-yield properties of the bone tissue while the organic matrix alone responsible for the post-yield mechanical properties.

### 2.5.4 Strain-rate effects

If the stress strain diagrams of a material are different at low and fast loading rates, the material is considered as strain rate sensitive. In strain rate sensitive material, the stress versus strain characteristics are dependent on the rate of loading applied. Viscoelastic materials are strain rate sensitive where as steel, at room temperature, is regarded as not strain rate sensitive. Several investigations have been performed to characterize static and dynamic mechanical properties of trabecular and cortical bone. Many fractures are provoked by sudden falls or by impacts that necessitate the investigation of bone properties at high strain rates. Understanding the mechanical behaviour of bones up to failure is necessary in order to diagnose and prevent accidents and traumas [24].

Mechanical characterization of bone tissues at low and high strain rates has been performed by several authors [23][24][33][34]. McElhaney [34] studied stress-strain curves obtained for various strain rates, on human femurs in compression. The character of the curves changed from ductile, with about 1% plasticity at $10^{-3}$ s$^{-1}$, to brittle at 1500 s$^{-1}$. Essentially the same results on bovine bone in tension were found by Crowninshield [23], and are shown schematically in Figure 2.6. The work by Pithiou [24] on bovine femurals was of statistical nature. Both quasistatic with imposed displacement (speeds of 0.5, 5, 10 and 500 mm/min) and dynamic loading (velocity of 1 m/s) tests were performed. It was found that at quasistatic rates ($10^{-3}$ to 1 s$^{-1}$), bone was stronger than at the dynamic rate of 120 s$^{-1}$. Another study revealed that human cortical bone was ductile at 0.1 s$^{-1}$ and became more and more brittle as the rate was increased to 20 s$^{-1}$ [35].

Natali [29] developed a visco-elasto-plastic constitutive model to investigate the mechanical behaviour of the human cortical bone tissue, accounting for anisotropic properties and post-elastic and time-dependent phenomena. The finite element model of the specimens was developed with the use of experimental data obtained from literature on the behaviour of cortical bone obtained from bone samples. Numerical simulations of the experimental tests were implemented for constant strain rates. A comparison of the numerical results and experimental data is given in Figure 2.7 showing an increase in the yield stress of the bone tissue with strain rate.
Figure 2.6 In the plastic region a tenfold increase of the strain rate causes an increase of about 25 MPa in the flow stress (After [23])

Figure 2.7 Experimental data and model results for constant strain rate tensile tests [29]

2.5.5 Bone nanoindentation

Nanoindentation is widely used by the materials science community to study the mechanical properties of thin films, small structures and other microstructural features. This technique offers a means by which intrinsic mechanical properties of the microstructural components can be measured directly. A typical load-displacement curve obtained from nanonindentation test is shown in Figure 2.8. As the load is increased, the indenter sinks into the material deforming it both elastically and plastically. When the indenter is unloaded, the material recovers by a process that is primarily elastic.

Rho [36] used the nanoindentation method to investigate elastic properties of several of the microstructural components of human vertebral trabecular bone and tibial cortical
bone. The specimens excised from tibia were tested in the longitudinal direction, yielding moduli of 22.5±1.3 GPa for the osteons and 25.8 ± 0.7 GPa for the interstitial lamellae. The obtained results showed a significant difference in the Young’s moduli of interstitial lamellae and osteons in the bone tissue, with the elastic modulus of the interstitial lamellae being about 15% greater. The elastic modulus of a macroscopic sample of cortical bone was assumed to fall somewhere between the interstitial lamellae and the osteons, with the exact value related to the microstructural arrangement and the of the constituents and relative volume fractions. The effects of elastic anisotropy on nanoindentation measurements was investigated for human tibial cortical bone [37]. Nanoindentation tests were conducted in 12 different directions in three principal planes for osteonal and interstitial lamellae. The measured indentation moduli were found to be dependent on the indentation direction and revealed anisotropy. A typical load-displacement graph obtained from nanoindentation test is shown in Figure 2.8.

**Figure 2.8** A schematic illustration of typical indentation load-displacement behaviour during one cycle of loading and unloading, showing quantities used to determine hardness and Young’s modulus from the data. The quantities are the peak indentation load \( P_{\text{max}} \), the maximum indenter displacement \( h_{\text{max}} \), and the initial unloading stiffness \( S \) (After [36])

### 2.5.6 Microhardness measurements

Another technique used since the early experimental studies on bone tissues is the use of microhardness testing since the method is sensitive to the in-plane symmetry of the surface under tests. Nanoindentation and microhardness indenters (the Vickers symmetric tip or the knife-edge shaped Knoop indenter) have been used in these efforts.

The Knoop hardness number \((KHN)\) is the ratio of the load applied to the indenter, unrecovered projected area \(A\) (mm²):
\[ KHN = \frac{P}{A} = \frac{P}{2CL}, \]  
(2.3)

Where \( P \) is the applied load in kgf, \( L \) is the measured length of long the diagonal of indentation in mm, \( C = 0.07028 \) is constant of the indenter relating the projected area of the indentation to the square of the length of the long diagonal.

The Vickers Diamond Pyramid harness number is the applied load (in kgf) divided by the surface area of the indentation \( (\text{mm}^2) \):

\[ HV = \frac{2F \sin 136^\circ}{d^2} \]  
(2.4)

Where \( F \) is the load, \( d \) is arithmetic mean of the two diagonals.

The microhardness properties of the structures of parallel-fibred and lamellar bones up to the tens of micrometers scale was studied by Ziv [38]. Vickers microhardness measurements were made and the microhardness values of parallel-fibered bone, examined in three orthogonal directions were found to be different. The average measured values were 0.480 GPa, 0.598 GPa and 0.706 GPa respectively in the periosteal, longitudinal, and transverse directions. Broz [39] used timed immersion in buffered ethylenediaminetetraacetic acid (EDTA) to change the mineral content at each level in the structural hierarchy of the cortical bone. The microhardness measurements were obtained with a 136° pyramid-shaped (Vickers) diamond indenter. A 50 g load was applied to fresh bovine bone samples for at least 10 s. The mineralization and site-specific properties of the mineralized bone samples were studied.

\[ \text{Figure 2.9} \text{ Means and standard deviations of hardness obtained from nanoindentation tests} \text{ [40]} \]
cores were not significantly affected by buffered EDTA immersion. Zysset [40] measured the hardness of wet human femur (cortical) at the lamellar level of organization. Specimens were retrieved from the diaphysis and the neck of eight human femurs. Results of previous studies on hardness were extended and significant differences in hardness have been found for various tissue types, anatomical sites and individuals as shown in Figure 2.9.

2.5.7 Fracture properties

The concepts of fracture mechanics are widely applied for in-depth analysis of the hierarchical structures biological tissues such as bone, tooth and shells under various kinds of loadings. Studies related to the bone’s fracture mechanics have characterized it resistance to fracture at the onset of a crack using a critical stress intensity factor or a critical strain energy release rate.

**FE study of bone fracture**

Finite element studies related to bone tissues are mainly concerned with a composite model. A cortical bone-like composite material was modelled in [18] using circular inclusions as osteons and their cement lines. A two-inclusion model was developed to investigate how the crack location, inclusion size and property change affect the extension of an initial crack between the inclusions. Finite element simulations were performed to compute stress distributions and strain energy release rates. For the two-inclusion model, mode I was predominant and the stress intensity factor (SIF) $K_I$ was used to determine crack stability. A crack was modelled as a line attached to the corresponding model. The four-inclusion model (see Figure 2.10) was optimized in order to arrest any crack

**Figure 2.10** Four-inclusion model with initial horizontal crack located at the centre of four inclusions. 2l is crack length [18]
initiated within the interstitial material between the inclusions. The results demonstrated that inclusions with effective elastic moduli smaller than that of the interstitial material always tended to attract cracks. On the other hand, an inclusion with an effective modulus greater than that of the interstitial material showed a tendency to repel the cracks. It was shown that crack propagation direction can be changed by varying the inclusion’s sizes and elastic properties.

Budyn [41] analysed cortical bone for multiple crack growth in a unit cell under tension using eXtended finite element method. The Haversian microstructure was modeled by a four-phase composite: The Haversian canal, osteons, cement line and matrix was discredited by FEM. The geometric and mechanical bone parameters were obtained by nanoindentation measurements and were randomly distributed to mimic the nature of bone at the micro scale. Cracks were initiated at the micro scale at osteonal locations where a critical elastic-plastic strain driven criterion was met. The cracks were randomly located within the microstructure as shown in Figure 2.11. SIFs were computed at each crack tip and a load parameter was adjusted so that the stress intensity factors remain at the critical value. The cracks were then grown in heterogeneous linear elastic media when a critical stress intensity factor criterion was met. It was concluded that cement lines appear as critical elements in the protection of bone against fracture and relevant structural elements to study their evolution in osteoporosis.

Figure 2.11 (a) Initial crack paths, (b) final crack paths at step 55 (broken microstructure). The ratio of cement line average SIFs to osteonal average SIFs is about 3: (c) final crack paths at step 88. The ratio of cement line average SIFs to osteonal average SIFs is about 6 [41]

Experimental study of bone fracture

Elastic-plastic fracture mechanics was applied to study bone's fracture toughness [42]. The $J$ integral was used to quantify the total energy spent before bone fracture was
initiated. Twenty cortical bone specimens were cut from the mid-diaphysis of bovine femurs. Ten of them were prepared to undergo transverse fracture and the other 10 were prepared to undergo longitudinal fracture (see Figure 2.12). The test specimens were cut to the depth of the starter notches \(a_0\) equal to 2 mm, which was half the height of the specimens \(W\) and thickness \(B\). Single-edge notched beam specimens were cut from the mid-section of a bovine femur.

**Figure 2.12** Preparation of notched specimens. Starter notches were cut in the middle of each specimen. SEM images of the notch geometry and notch tip are also shown. Scale bar in notch tip inset is 20 \(\mu\)m [42]

After fracture, the \(J\) integral of each specimen was estimated using the following relations [42]

\[
J_{\text{total}} = J_e + J_p = \frac{K_e^2}{E'} + \frac{2}{B_N b_0} A_{pl},
\]

(2.5)

Where \(J_e\) is the \(J\) integral of the elastic deformation, \(J_p\) is the \(J\) integral of the plastic deformation, \(K_e\) is the critical stress intensity factor, \(E'\) is the elastic modulus for plane stress condition. \(B_N = B\) (total depth of the two side grooves), \(b_0\) is the height of the uncracked ligament, and \(A_{pl}\) is the area of the plastic deformation part in the load-displacement curve. The critical stress intensity factor was calculated as

\[
K_e = \frac{PS}{B_N} \left(\frac{1}{3}\right)^{\frac{3}{2}} \frac{f(\alpha)}{W^{\frac{3}{2}}},
\]

(2.6)

Where \(P\) is the maximum load, \(S\) is the support span of the bending test and \(\alpha = \frac{a_0}{W}\) and
\[
f(\alpha) = \frac{\frac{1}{3\alpha^2}[1.99 - \alpha(1 - \alpha)(2.15 - 3.93\alpha + 2.7\alpha^2)]}{2(1 + 2\alpha)(1 - \alpha)^{\frac{1}{2}}}.
\]  

(2.7)

From the obtained \( J_{\text{total}} \), linear elastic equivalent fracture toughness, \( K_{\text{lc}} \), of the bovine specimens can be calculated:

\[
K_{\text{lc}} = \sqrt{J_{\text{total}} E'}.
\]  

(2.8)

The \( J \) integral of the transverse-fractured tests specimens was measured to be 6.6 kPa m, which was 187% greater in magnitude than that of the longitudinal fractured specimens (2.3 kPa m) (Table 2.2). The energy in the plastic deformation of the longitudinal-fractured and transverse-fractured bovine specimens was measured to be 3.6–4.1 times the energy spent in the elastic deformation. The study showed that the toughness of bone calculated with \( J \) integral was greater than the that measured using the critical stress intensity factor.

**Table 2.2** Fracture properties of bovine femur [42]

A crack propagation-based approach to measure bone toughness was compared with crack initiation approach by Vashishth [43]. Analyses of data from previously tested bovine and antler cortical bone compact specimens demonstrated that, in contrast to crack initiation approach, the crack propagation approach successfully identifies the superior toughness properties of red deer’s antler cortical bone over a bovine cortical bone (see Figure 2.13).
Figure 2.13 (a) Average values of initiation fracture toughness (MPa) for antler and bovine cortical bone; (b) average slope of crack growth resistance or R-curves for antler and bovine cortical bone. Note that a greater slope value for antler bone indicates that it displays a steeper increase in the crack growth resistance with crack extension than bovine bone [43].

The critical energy release rate of human tibial and femoral mid-diaphysis was measured for various crack propagation directions with three-point-bending tests using controlled crack extension [44]. The local structure was found using small-angle X-ray scattering SEM and polarised light microscopy and was linked to the energy needed for crack propagation (Figure 2.14). It was found that the collagen angle was responsible for changing the fracture behaviour of bone from brittle to quasi-ductile. A significant increase in the critical energy release rate as well as a change of the appearance of the crack path from straight to deflected was observed. It was straight and smooth for crack propagation in longitudinal direction with a low value for the energy to extend the crack and

Figure 2.14 (a) Human cortical bone specimen with orientation; (b) appearance of crack paths as well as level of the $J$-integral for two orientations of mineral platelets determined by small-angle X-ray scattering (SAXS) [44]
strongly deflected with a much higher value for circumferential and radial direction as shown in Figure 2.14. The different behaviour of the crack path was visualised by the SEM-micrographs. The crack direction was seen to be influenced by the direction of the fibrils with respect to the loading axis.

2.5.8 Density

The density of the bone can be defined either at the tissue or material level. The difference between the two arises from the porosity and the presence of certain structural features such as osteocyte lacunae, canaliculi, osteonal cavities. At the tissue level, the ratio of wet mineralised mass to the volume occupied by the tissue is known as the ‘apparent density’. At the material level, the mineralised mass over the volume occupied by the material itself is called ‘material density’. A more straightforward concept to represent bone’s porosity is BV/TV (ratio of bone material volume over tissue volume).

A comprehensive study was recently conducted to investigate the relationship between the aforementioned quantities in cortical and trabecular bone of an elephant using quantitative computer tomography (QCT) and absorptiometry methods [45]. In that study the specimen was extracted from right femur of an adult elephant of 24 years old. The study found an exponential increase of elastic modulus with an increase in the described variables and is shown in Figure 2.15.

Figure 2.15 Variation of elastic modulus of bone with its apparent density and material density [45]

Huiskes [46] summarised the results of three investigators who measured the density of cortical bone. The range was $1.86 \times 10^3 \text{ kg/m}^3$ to $2.9 \times 10^3 \text{ kg/m}^3$ with an average of $2.2 \times 10^3 \text{ kg/m}^3$. 
2.6 Comparison of Properties of Human and Bovine Cortical Bone

There is a large amount of data in the literature on the mechanical properties of human and bovine cortical bone. Table 2.3 compares the mechanical properties of human and bovine cortical bone. All the data in Table 2.3 are from static tests on wet bone.

Table 2.3 Comparison of mechanical properties of human and bovine cortical bone

<table>
<thead>
<tr>
<th>Property</th>
<th>Bone Type</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Human</td>
</tr>
<tr>
<td>Density, $\rho$ (kg/m$^3$)</td>
<td>1800–2000</td>
</tr>
<tr>
<td>Young’s modulus, $E$ (GPa)</td>
<td>10–17</td>
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<td>Shear modulus, $G$ (MPa)</td>
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<td>Poisson’s ratio</td>
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<td>Compressive strength, $\sigma_p$ (MPa)</td>
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</tr>
<tr>
<td>Tensile yield strength, $\sigma_y$ (MPa)</td>
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</tr>
<tr>
<td>Toughness, $G_{tt}$ (J/m$^2$)</td>
<td>350</td>
</tr>
<tr>
<td>Fracture toughness $K_{tc}$ (MN/m$^{3/2}$)</td>
<td>3</td>
</tr>
</tbody>
</table>

2.7 Thermal Properties of Bone

Much has been written about the mechanical properties of bone. Characterisation of bone mechanical properties is essential in understanding the influence of the mechanical environment around bones and joints for therapeutic effect. Bone is a complex biological tissue also in terms of its thermal properties. The measurement of thermal properties of bone is important to calculate the temperature rise in bone in a cutting process using analytical and numerical modelling. Despite the fact that all orthopaedic interventions induce significant thermal effects, the thermal properties of bone are less well documented than mechanical ones. The heat capacity and thermal conductivity of bone are important factors affecting a temperature rise during the cutting operation. These factors have an inverse relationship with the temperature rise. The lower the values, the higher the temperatures generated in the material being cut. When drilling metals the metal chips carry away nearly 85% or more of the heat generated [47]. The temperature of the bone will rise more during the drilling process due to its poor thermal heat transport and chips carrying away a smaller percentage of the heat compared to other engineering materials of higher thermal conductivity and heat capacity such as metals [47]. The heat generated during bone drilling is due to its lower value of thermal conductivity compared to that of metals.
Bone can be exposed to heat from several types of sources, including drilling [48], laser ablation [49], and the curing of cements used in hip arthroplasty [46]. The elevated temperatures, which result from such exposure, may lead to cell death, i.e., to thermal necrosis, which in turn can lead to infection and reduced mechanical strength [50]. The extent of necrosis depends on temperature rise and duration. The equations of unsteady heat conduction can be used to calculate temperature distribution in bone provided the geometry, heat input, and thermal properties of bone are known. The knowledge of thermal properties such as thermal conductivity, specific heat and coefficient of thermal expansion are essential in any work devoted to the goal of understanding and reducing thermal necrosis during cutting procedures in which heat is produced.

2.7.1 Thermal conductivity

Depending on the age, bone type, anatomical direction and experimental conditions, a wide range of thermal conductivity magnitudes has been reported for bone tissue in the literature ranging from about 0.2 W/m K to around 13 W/m K. Thermal conductivity of equine cortical bone was measured using inverse conduction procedure in conjunction with a guarded hot plate device by Moses [51] and the results were compared to those reported in the literature. Results of the tests showed values of thermal conductivity of 0.7 W/m K for dry test specimens and 0.80 W/m K for saturated specimens.

Thermal conductivity, under load and at low temperatures using a simple calorimeter was measured by Wifp [52]. Cylindrical specimens of 0.625 cm diameter and 0.94 cm length were cut from cortical bone. The thermal conductivity at 20 K was measured as 53±3 mW/m K. In another study by Davidson [53], thermal conductivity and its variation

Figure 2.16 (a) Schematic drawing of a cross-section of bovine femur, showing the location and shape of a typical specimen; (b) schematic cross-section of the experimental apparatus used to measure thermal conductivity of cortical bone. A, Plexiglas block; B, bone specimen; C,D, aluminum plates; E, heating element; F, water bath; G, support wall; H, insulation; I, guide rods; J, thermocouples [53]
with the anatomical direction of fresh bovine cortical bone was measured. The experimental apparatus, illustrated schematically in Figure 2.16, was designed to create a one-dimensional heat flow through bone specimen. The conductivity was found to be 0.58±0.018 W/m K in the longitudinal direction, 0.53±0.030 W/m K in the circumferential direction and 0.54±0.020 W/m K in the radial direction. Because the directional differences were small, it was concluded that bovine cortical bone can be treated as thermally isotropic.

The range of thermal conductivity reported in literature are summarised in Table 2.4.

**Table 2.4** Summary of thermal conductivity of cortical bone reported in literature.

<table>
<thead>
<tr>
<th>Reference</th>
<th>Type of bone</th>
<th>Thermal conductivity (W/m K)</th>
</tr>
</thead>
<tbody>
<tr>
<td>[51]</td>
<td>equine femur</td>
<td>0.69 – 0.88</td>
</tr>
<tr>
<td>[52]</td>
<td>cow Metacarpal</td>
<td>0.53</td>
</tr>
<tr>
<td>[53]</td>
<td>bovine femur</td>
<td>0.53 – 0.58</td>
</tr>
<tr>
<td>[54]</td>
<td>human</td>
<td>3.5</td>
</tr>
<tr>
<td></td>
<td>elephant</td>
<td>4.3</td>
</tr>
<tr>
<td></td>
<td>ox</td>
<td>5</td>
</tr>
<tr>
<td>[55]</td>
<td>human</td>
<td>0.38</td>
</tr>
</tbody>
</table>

**2.7.2 Specific heat**

Specific heat of a material is effectively a measure of how easily it “heats up”. The amount of heat developed within a bone tissue depends largely on specific heat capacity. The rate of heat generation during bone cutting depends on various parameters of the cutting process, affecting the temperature at the tool-bone interface. The maximum temperature a specimen develops depends on the balance between heat gain and heat loss. When the heat gain and heat loss are equal, the specimen is in a state of equilibrium. In a state of equilibrium, the temperature of a specimen depends on the exposure temperature and thermal conductivity. In clinical practices, temperature equilibrium may not establish when cutting bones with a high-speed cutting tool such as a drill, since the exposure times are usually too short. In a non-equilibrium state, the temperature reached for short exposure time is a function of the exposure temperature and time, for a constant value of specific heat capacity [54].

An axisymmetric model of the tibia was applied within the framework of the finite element method and analysed the heat conduction from bone cement in total knee arthroplasty (TKA) using numerical simulation by Fukushima [55]. The study hypothesised
the thermal necrotic map of bone and suggested a method for preventing thermal necrosis of bone. The specific heat value used in simulations for cortical bone was 1.26 J/kg °C.

Huiskes [46] summarised the published values for the specific heat of bone, with one researcher measured values in a range from $1.15 \times 10^3$ kJ/kg K to $1.73 \times 10^3$ kJ/kg K and two reported the same value $1.26 \times 10^3$ kJ/kg K.
Chapter 3. Mechanics of Bone Cutting

3.1 Introduction

Bone is cut in orthopaedic surgery for pin insertion and plate fixation when fractured or damaged by diseases (e.g. Knee orthoplasty, osteoarthritis). In 2000, about 12 percent of the total population in United States, more than 15 percent in European countries and approximately 19 percent in Japan are reported to live over the age of sixty-five. Due to an extensive increase in human aging, associated disease and demand for optimised bone cutting methods, it is necessary to understand bone cutting mechanics in its entirety. In orthopaedics, the cutting is either with a tool having rotational motion (drilling, milling) or linear motion (plane cutting with chisel). The mechanical operations performed on bone such as drilling, milling, sawing and cutting with a chisel are similar to those performed on other materials. Among mechanical operations on bone, drilling is the most common and frequent in orthopaedic surgery. This chapter mainly focuses on the experimental techniques employed in measuring mechanical and thermal responses of bone tissues during a cutting process. As the most frequently used tool in orthopaedics surgical procedures is a mechanical drill, more emphasis is placed on drilling process. In this chapter, analysis of stresses and forces are mainly based on the work presented by Shaw [56]. In the first section basic mechanics of the cutting process is discussed.

3.2 Mechanics of Cutting

The basic mechanisms of cutting processes depend on the direction of the tool relative to the work piece. In the general form the tool is oriented perpendicular to the work piece and the process is called orthogonal cutting (see Figure 3.1). Mechanics of this type of cutting has been extensively researched and developed by the metal cutting industry over the past 100 years and has been successfully applied to bone cutting. This cutting condition involves two major forces, the normal force and the thrust or feed force. The resultant force can be derived from vector resolution of the normal force and the thrust force. This operation has been defined by Merchant [57][58] as orthogonal cutting (Figure 3.1). The performance of the cutting process is related to the amount of work required to cut the material as defined in previous reserach [57][58][59]. This analysis has been reported to be valid for the cutting of bone [60][61][62].

In material cutting, large deformations take place in the vicinity of a tool-work piece interaction zone, and cutting conditions affect considerably strain rates and temperatures. Thermomechanics of the tool-work piece interaction, which is important for a regime with
multiple micro impacts in the process zone as in the case of ultrasonically-assisted cutting and other specific features of the cutting process cannot be fully estimated from analytical solutions and assessed by experimental methods. FE modelling is a main computational tool for simulation of the cutting process and of tool-work piece interaction in bone cutting that can provide detailed information on forces, stress, temperature, chip shape and also can predict a set of parameters for an efficient cutting process.

3.2.1 Chip formation

Chip formation requires concentrated shear along a distinct shear plane. The deformation of material starts after it reaches the primary shear zone (the deformation zone which stretches from the tool tip to the free surface of the work piece). The material then undergoes a substantial amount of simple shear as it crosses this thin zone. There is essentially no further plastic flow as the chip proceeds along the tool face. Uncut chip thickness $t_1$ is closely related to feed, but is used from the point of view of the chip formation process. The chip formation involves separation of the work piece material in the vicinity of the cutting edge and its subsequent shearing in the primary shear zone near the shear plane (line OD in Figure 3.1). Large deformation takes place in this region at very high strain rates, and for some materials and under certain cutting conditions this leads to catastrophic shear and fracture resulting in discontinuous chips. Such fragmented chips are one of the principal classes of chip shape. However, under the majority of cutting conditions, ductile materials do not fracture on the shear plane and a continuous chip is produced.

Figure 3.1 illustrates the formation of a chip of thickness $t_2$ from an undeformed layer of thickness $t_1$ by the action of a tool with rake angle $\alpha$. Two new surfaces are formed in this process, the new surface of the work piece OA and that of the chip OC. The main deformation region is situated in the zone around line OD going from the tool edge to the position where the chip leaves the work piece surface and is known as a primary deformation zone. Hence, for the purpose of simple analysis, the chip is assumed to be formed by shear along the shear plane OD. A secondary deformation zone is observed along the rake face of the tool, where a seizure often occurs at the interface between the chip and tool along a contact length $L_c$ (OB in Figure 3.1). The shear plane OD is inclined to the shear angle $\varphi$ to the cutting direction OA that is an important parameter in chip formation mechanics, and various models are employed in order to predict it.
3.2.2 Forces involved in chip formation

In a free body diagram of orthogonal cutting, two types of forces are considered in the chip – the force between the tool face and the generated chip (\( R \)) and the force acting along the shear plane (\( R' \)). These forces are equal when an equilibrium condition is developed. The forces \( R \) and \( R' \) are may be resolved into three of components (Figure 3.2): (1) the vertical and horizontal components, \( F_Q \) and \( F_P \); (2) perpendicular and along the shear plane, \( N_S \) and \( F_S \); (3) along and perpendicular to the tool face, \( F_C \) and \( N_C \).

Figure 3.2 Free body diagram of chip [56]

A compact diagram may be formed if the forces \( R' \) and \( R \) are plotted at the tool point instead of their actual points of application along the shear plane and tool face. The forces \( R \) and \( R' \) (which are equal in magnitude and parallel) are coincident making the diameter of the dotted reference circle shown in Figure 3.3. The line terminating at the ends of diameter \( R \) and intersect at a point on the circle form right angles, there is a convenient mean for graphically resolving \( R \) into orthogonal components in any direction.
Figure 3.3 Composite cutting force circle [56]

The shear and friction components of the force in terms of the horizontal and vertical components \(F_p\) and \(F_q\) may be combined to form analytical relationships. These components are the generally measured experimentally with a dynamometer. From Figure 3.4 it is evident that

\[
F_s = F_p \cos \varphi - F_q \sin \varphi, \tag{3.1}
\]
\[
N_s = F_q \cos \varphi + F_p \sin \varphi = F_s \tan(\varphi + \beta - \alpha). \tag{3.2}
\]

Similarly

\[
F_c = F_p \sin \alpha + F_q \cos \alpha, \tag{3.3}
\]
\[
N_c = F_p \cos \alpha - F_q \sin \alpha. \tag{3.4}
\]

The components of the force at the tool face are of importance as they enable the coefficient of friction on the tool face \(\mu = \tan \beta\), where \(\beta\) is the friction angle (shown in Figure 3.4):

\[
\mu = \frac{F_c}{N_c} = \frac{F_p \sin \alpha + F_q \cos \alpha}{F_p \cos \alpha - F_q \sin \alpha} = \frac{F_q + F_p \tan \alpha}{F_p - F_q \tan \alpha}. \tag{3.5}
\]

Figure 3.4 Resolution of resultant cutting forces: (a) Shear Plane; (b) tool face [56]
3.2.3 Drill geometry and design

Orthopaedic surgical operations involve drilling and tapping for the insertion of screws or pins into bone. Technological improvements in design elements of a modern surgical drill allow its penetration into bone in an efficient manner producing channels of uniform size. Cutting of a solid material requires a blade to engage it at a critical angle in order to shear a layer off the material. The cutting process depends on the material properties of the blade and the material being cut. All cutting tools use a blade edge that cut a material surface. The shape of the blade is determined by various angles [63] (Figure 3.5):

**Rake angle:** the angle at which the cutting face is presented to the material.

**Clearance angle:** the angle by which the flank (non-cutting portion of the blade) clears the material.

**Wedge angle:** the angle between the cutting face and the flank.

**Point angle:** the angle between the two cutting edges of a drill.

![Figure 3.5](image) Simplified diagram of cutting edge with important design features [63]

A twist drill is a complex cutting tool. To understand the function of its elements, the geometry of a drill can be broken down to different elements (Figure 3.6 to Figure 3.9).

**Chisel edge:** the edge at the end of the web connecting the cutting lips of a drill.

**Flutes:** helical grooves produced in the body of the drill for chip removal.

**Land:** the peripheral location on the drill body between mating flutes.

**Lips:** the cutting edges of the drill connecting the chisel edge to the periphery.

**Chisel edge angle:** The angle between the chisel edge and the cutting lips.

**Helix angle:** the angle between the land and plane containing the longitudinal axis of the drill.

**Point angle:** the angle between the cutting lips projected upon a plane parallel to the drill axis and parallel to the two cutting lips.
The flute of a drill can be defined by its helix angle which may be slow, standard or quick as shown in Figure 3.9. The angle represents the number of turns per unit length. A slow helix is used in drilling materials where easy evacuation of chips is desirable. For fast evacuation of chips to avoid unnecessary blocking of flutes, fast helix is required. The length of the flute affects the evacuation of chips from deep holes. The process of drilling requires a twist drill to penetrate into the material and cut portion of it through the shear process as it advances. The chisel edge located at the centre does not perform cutting of the material and only compresses the material ahead of it. The cutting lips shear the material surface and produce plastic deformation along the shear planes. As a result of the drilling
process, failure of the material occurs near the sharp cutting lips of the drill generating a chip that flows in the flutes.

**Figure 3.9** Drawing showing various helix angle and worm spiral pattern flutes [64]

The components of forces acting on the cutting lip during drilling is shown in Figure 3.10. The resultant cutting force $R$ can be resolved into three orthogonal directions, $X$, $Y$ and $Z$. $P_x$ is the principal cutting force in the direction of cutting velocity and is responsible for producing drilling torque. $P_x$ is the thrust force at the cutting edge acting in the feed direction. $P_y$ is a radial force which is responsible for balancing a similar force acting on the other lip. During drilling operation, the chisel edge produces extrusion effect which gives rise to a considerable magnitude of thrust force [65]. The total thrust force of

**Figure 3.10** Diagram illustrating the force vectors acting at the tip of a drill-bit during drilling [63]
a drill comprises of $P_X$, the component of the force at the cutting location and extrusion effect at the chisel edge with frictional effects. The chisel edge where the cutting velocity is low and the rake angle highly negative, greatly increases the thrust during drilling [65].

### 3.3 Orthopaedic Surgical Tools and Materials

The design of bone surgical tools, as we know them today, dates back to the 17th or 18th century [66]. Orthopaedic surgical tools such as drills, chisels, toothed saws and knives are similar to those used in the wood industry with few technical enhancements. Each surgical tool with different design, cut the bone differently thus affecting mechanics of the process. In addition to the tool design, its surface topology and material (composition) is important for bone cutting since it affects the heat transport across the tool-bone interface.

Implants used in orthopaedic surgery must be made from corrosion-resistant materials. The employed steels are mainly composed of 0.3% carbon, 17-19% chromium, 13-15% nickel, 3% molybdenum, 2% manganese and up to 60% iron, with other trace elements [67]. Plates and screws used in bone implant surgery are made from titanium for its better corrosion-resistance and biocompatibility. The implants made from titanium can remain in the human body for long periods of time with little or no allergic reaction to the adjacent tissues. Standards ISO(5832-1), AISI (316L) and DIN (17-440) deal with implant materials.

The steel material of the drills used in orthopaedic surgeries are not necessary to be as corrosion-resistant as implant materials, since they are in contact with bone tissue for short periods of time [67]. More important characteristics of these tools are the retaining of the sharpness of the cutting edges and withstanding of repeated sterilisation cycles in an autoclave at temperatures of up to 1358°C. Orthopaedic surgical tools are manufactured to standards (DIN 1.4112 or AISI 440B) and have chemical compositions of approximately 0.85% C, 18% Cr, 1% Mo, 1% Mn 1% Si, and trace elements, with the remainder made up of iron [67]. The ultrasonic scalpel used in maxillofacial surgery can be found in Ref. [68]

### 3.4 Thermal Necrosis of Bone

Necrosis or thermal injury means bone death and can severely delay the healing process of an infected area after surgery. Necrosis may cause problems for implantation of surgical screws and pins. If necrosis sets in, it can break down the bone around the implantation site, causing the loosening of fixtures (screws, pins) and can dramatically weaken the whole structure [69]. The causes of thermal necrosis are still under study and
could be due to the denaturing of bone alkaline phosphatase at a critical temperature, and for a specific time duration with regard to the severity of the temperature induced. The temperature and time duration necessary for thermal necrosis to occur depends mainly on the type of bone.

One of the most thorough investigations of threshold temperature was done by Moritz and Henriques [70]. They measured the amount of time required to produce damage to the dermal and epidermal layers of both human and porcine skin over a large temperature range (44°C to 100°C). They established, for instance, that a temperature of 70°C will kill the epithelial cells immediately, and a temperature of 50°C will do so after 30 seconds. Similarly, temperature of 45°C would have to be maintained for more than five hours to harm the cells. The study found that as the temperature increased, the amount of time required to initiate thermal necrosis decreased, resulting in a time-temperature curve similar to that shown in Figure 3.11.

![Figure 3.11](image-url)  
**Figure 3.11** Time-temperature curve for thermal necrosis of epithelial cells (the graph is based on data from [70])

The time-temperature dependence of thermal damage can be expressed by the Arrhenius relationship presented using equation 3.6 [71]:

$$\Omega = \int_{0}^{t} Ae^{\frac{-E_{a}}{R(T+273)}} \, dt,$$

where $\Omega$ represents the thermal injury incurred by the bone tissue from time 0 to time $t$, $A$ is referred to as the frequency factor (s$^{-1}$), $E_{a}$ is activation energy (J/mole) and $R$ is the universal gas constant. The Arrhenius relationship presented above was derived from the
data collected by Moritz [70]. The Arrhenius parameters presented by Henriques [71] for epithelial tissue were $A = 3.1 \times 10^{98}$ s$^{-1}$ and $E_a = 6.27 \times 10^5$ J/mole. It was determined that if the thermal injury incurred by epithelial tissue, $\Omega$, exceeded a value of 0.53, then thermal necrosis would ensue [71].

Lundskog [54] described extensive necrosis in cortical bone at a temperature of 70°C and measured a threshold temperature of 50°C at 30 seconds exposure. Eriksson [72] discovered cortical necrosis in rabbit bones subjected to 47°C for 1 minute, whereas there have been reports of necrosis in dog femoral bone at 50°C. The results by Eriksson [72] were consistent with those of Moritz [70]. It is assumed that on average the threshold temperature, at which necrosis is most likely to appear is around 50°C–60°C, but the time duration is dependent on the severity of temperature, according to the variety of published experimental results. Frolke [73] has suggested that temperatures of $\leq 77$°C during reaming would be safe for bone tissue with short-term exposure.

Measures to reduce the likelihood of temperature increases were investigated in the past. The results have shown wide agreement that a sharp, coolant irrigated cutting tool effectively controls excessive drilling temperatures. But there is still disagreement over how the drill geometry, force and speed affects temperature in the cut bone. To sum up various findings, it is evident that the subject of bone necrosis is a complex one, and the debate continues on how the results of such experiments should be reflected in clinical practice.

### 3.5 Bone Drilling

Drilling of bones for orthopaedic and dental purposes has been a common practice for centuries. It is a major part of bone surgery during preparation for insertion of a fixative orthopaedic implant such as a nail, screw or wire or before insertion of a bone graft to enhance bone healing [74]. A typical application of drilling in femoral fracture surgery is shown in Figure 3.12.

Presently, mechanical rotary drills are the main type used in the clinical setting. Rotary drilling performed at a wide range of speeds, from low to moderate (<10000 rpm) up to ultra-high (>15000 rpm) [75]. Other bone drilling methods such as laser drilling are briefly discussed in [76][77]. The major concern in those studies was the bone tissue’s thermal damage, which occurred when the temperature rise was above the tissue’s threshold value. The drilling process may cause overheating of the bone tissue that can decrease the moisture contents (increase dryness) around the hole. Dryness of human cortical bone has been shown to increase tensile and compressive strength and modulus but decrease tensile
and compressive strength [Evans-1976]. Previous research was focused mainly on
avoiding thermal damage by changing various cutting parameters like forces, drill-bit
geometry, the feed rate, drilling speed, and cutting tool geometry and material etc.

3.5.1 Drilling temperatures vs. drilling force

Bone drilling operation and measurements of generated cutting forces and temperatures
have been the interest of researchers for many decades. Like drilling of other materials,
bone drilling requires a force to push the tool into the material. In bone surgery, a surgeon
needs to apply a certain amount of force, which controls the drill but does not harm the soft
tissue around. One of the ways to achieve this is to use robot assisted surgery, where a
sensitive force feed-back system controls the tool action.

Studies have found that maximum cortical bone temperatures are strongly correlated to
the drilling force [78]. Previous studies have demonstrated how the drill penetration force
affected bone temperatures [79][80][81][82]. The magnitude of drilling forces reported in
the literature is given in Table 3.1

<table>
<thead>
<tr>
<th>Reference</th>
<th>Forces (N)</th>
<th>Drill Speed (rpm)</th>
<th>Drill Diameter (mm)</th>
<th>Type of Bone</th>
</tr>
</thead>
<tbody>
<tr>
<td>[79]</td>
<td>20, 59, 118</td>
<td>345, 885, 2900</td>
<td>3.2 surgical bit</td>
<td>Human femur</td>
</tr>
<tr>
<td>[80]</td>
<td>1.5 – 9</td>
<td>up to 100000</td>
<td>2.5 surgical bit</td>
<td>Bovine femur</td>
</tr>
<tr>
<td>[81]</td>
<td>1.5, 4, 12, 24</td>
<td>1800, 2400</td>
<td>3.25 dental burr</td>
<td>Bovine femur</td>
</tr>
<tr>
<td>[82]</td>
<td>1.5 – 9</td>
<td>49000</td>
<td>2.5 surgical bit</td>
<td>Bovine femur</td>
</tr>
</tbody>
</table>
Some drilling experiments were performed on blocks of dentine, without irrigation in [83]. It was found that the amount of heat transferred to the specimen increased with the increasing load from 0.3 N to 0.5 N, then decreased as the load increased above 0.5 N. The heat-load behaviour was similar to the maximum temperature-load one studied by Abouzgia [82]. Although the free-running speed was held constant at 250,000 rpm throughout the experiment, the rotational speed was measured during drilling and found to decrease dramatically as the load increased. When Brisman [81] tested light drilling forces, cortical temperatures increased with the increase in forces from 1.5 N to 4 N. The study demonstrated that when using a lower drilling speed with forces of 12 N and 24 N, cortical bone temperatures increased with the force. However, some research showed cortical bone temperatures being inversely related to drilling forces. Abouzgia [80], for example, found that as forces increased from 1.5 N to 9.0 N in drilling bovine cortical bone, both maximum temperatures and their duration of application decreased.

Matthews [79] tested forces levels ranging from 20 N to 118 N in drilling human cortical bone and found that as the force increased, bone temperatures and their duration above 50°C decreased. The specimens were kept moist during the operation, but no irrigation was applied at the drill site. The effect of an increased drill speed was less pronounced but followed a similar trend as the effect of increased force. Although a clear reason for this temperature decrease was not known, it is possible that a larger force minimised the time required for penetration in the cortex of the bone. Another recent study on fresh frozen human cadaver femora also reported temperature decrease with an increase in drilling forces [78] (Figure 3.13). The study found that the increase in the drilling force up to 93 N resulted in decrease of the maximum temperature but above this load no

Figure 3.13 Cortical temperatures at 0.5 mm from the drill tract [78]
change was observed. The increase in the drilling force resulted in a significant decrease in the average duration of temperature elevations above 50°C.

3.5.2 Drilling temperatures vs. drilling speed

Much research can be found in the literature on the effect of drilling speed on other parameters such as cutting forces, the torque, surface finish and temperature for a number of materials ranging from metals to polymers, composites and biomaterials. The drilling parameters investigated most often have been the drill speed and the applied load or feed rate. The influence of these parameters has been measured in two ways: histological examination of the bone tissue, and measurement of the temperature rise at various distances from the drill site.

Histological response of bone tissues around a high-speed drill for the examination of thermal necrosis has been the focus of several studies. Necrotic tissue in the bone was found surrounding pins after they were inserted, without irrigation, into canine mandibles [69][84]. The thickness of the necrotic band was measured and was found to increased with drill speed from 125 to 2000 rpm. Moss [85] investigated the effect of higher rotational speeds, from 40,000 rpm to as high as 350,000 rpm on bone temperature. Holes were drilled, without irrigation, into canine mandibles with a number of different burs in three speed ranges and the acellular zone was measured around the defects after a period of two weeks. It was concluded that there was no disadvantage in using the ultra-high drill speeds. Matthews [79] studied the temperatures obtained when drilling human cortical bone, in order to determine the optimum drilling conditions for cortical bone using a standard drill. The experiment involved measuring the increase in temperature and its duration in the cortical wall of a human femur at different rotation speeds and pressures.

![Figure 3.14](image)

**Figure 3.14** Effect of drilling speed on average maximum temperature in cortical bone recorded at specific distances from drill [79]
applied to the drill (Figure 3.14). There were no significant differences in maximum temperature which could be attributed to the drill speed.

Brisman [81] observed that drilling at a speed of 1,800 rpm and at a load of 1.2 kg produced the same heat as the drill speed and load were increased to 2,400 rpm and 2.4 kg respectively. Independently increasing either the speed or the load resulted an increase in temperature in the bone. However, an increase in the speed together with the load allowed a more efficient cutting with no significant increase in temperature.

Abouzgia [80] used a surgical drill Stryker-100 in bovine bone drilling. The tests were conducted with 36 specimens at variable speeds (20,000 –100,000 rpm) and at different levels of the constant force (1.5–9 N). The results showed that the temperature rise and the duration for which the temperature remained in bone decreased with speed and force. The study suggested that drilling at high speed and with large load was more desirable than reported before. A recent study reported that heat generated during the drilling of human and bovine bone is proportional to the drilling speed [67]. A drill-bit with a 90° point angle was used for drilling 100 holes. Speeds of 800–1400 rpm were used when drilling with a 3.2 mm drill-bit to provide the best cutting conditions and maintaining temperatures at a manageable level. The experimental findings are given in Figure 3.15.

**Figure 3.15** Effect of speed and depth on temperature in drilling of bovine bone (a) and human bone (b) [67]

It is evident from the above studies that temperature rise in bone drilling is strongly related to various parameters i.e. speed, thrust force, feed rate, drilling depth and the type of bone under test. It is not simple to draw a final conclusion from these studies for optimal
parameters, since the results are not similar and in many cases are contradicting. This measurements reflecting ones close to the actual surgical operation.

### 3.5.3 Influence of drill geometry on bone drilling

The properties of bone vary with its state which determine its behaviour in drilling. In bone drilling, debris is mixed with blood and marrow fat causing it to stick to the drill flutes and clog them. In such situations quick helix flutes are more suitable [64] (see Figure 3.9). The chisel edge catches periosteum and eventually carries it to the flutes obstructing the flow of chips through the drill flutes as was shown by Jacobs [86]. The problem may be resolved by the use of split-point which imparts a positive rake angle and cutting action to the chisel edge [86]. It was the first study where seven different drill designs were tested ranging in speed from 100 rpm to 2360 rpm and with feed rates of 0.254, 0.508 and 1.27 mm/min. An optimum point angle of 90° was recommended despite the data showing that a 110° point angle produced smaller thrust force and torque. Modern surgical drills have been designed along these designs since.

Saha [65] developed a design with a 118° point angle and a split point flank with the clearance angle of 15–18°; the helix angle ranged from 34° to 36° depending on the drill diameter, and the flute geometry was parabolic. This design resulted in penetration at a faster rate with reduced thrust load than traditional designs. The peak temperature generated was also significantly lower. Wiggins [48] showed that a twist drill with a 118° point angle and a 28° helix angle required much lower torque per unit area of hole and energy per unit volume of bone drilled at a given feed rate, compared to a drill with a 60° point angle.

Natali [64] compared various commercial drill bits with surgical drills in the drilling of fresh human cadaver tibia. Five types of drills each of 2.5 mm diameter were used in temperature measurements: (1) a new standard orthopaedic bit (New AO drill), (2) an old (used) standard orthopaedic bit, (3) a standard orthopaedic bit, with its flutes deliberately blocked, (4) a high-speed steel twist bit, and (5) a split-point high-speed steel bit (the Black and Decker ‘Bullet’ drill bit). Figure 3.16 shows the graphs of the average temperature recordings for each drill made at 0.5 mm from the drill edge. The highest local temperatures were generated with the standard orthopaedic drill with artificially blocked flutes, and were sustained for longer. The new Stratec orthopaedic drill created less rise in temperature, but the lowest temperature changes were obtained with the split-point, fast helix drill (Black and Decker ‘Bullet’ drill). From the above discussion it can be concluded that certain optimal design parameters can be found for orthopaedic twist drills.
3.6 Ultrasonically-assisted Bone Cutting

The first use of ultrasonically-assisted cutting (UAC) of bone tissue was reported in 1950. Ultrasonic machining is already being utilised in metal cutting, food and other soft tissue cutting industries on a wide scale. Ultrasonically-assisted cutting of bones and other biological tissues particularly for removing the periosteum and coherent mass such as calcium are becoming popular in orthopaedic, neuro and dental surgery [1].

In recent in-vitro studies, the various factors that influence the cutting process of bovine bone by an ultrasonic chisel was analysed [9]. The experimental set up used in the experiments is shown in Figure 3.17. The relationship between the cutting rate and the
downward force that was generated when a bovine bone sample was cut with the ultrasonic chisel is shown in Figure 3.18. As the cutting rate increased, there was an increase in the downward force up to a feed rate of 40 mm/min. Then the force stayed nearly constant until the chisel reached a cutting rate greater than 56 mm/min when there was a fall in the force level. It was concluded that the process of bone cutting with the ultrasonic chisel requires low forces and cutting rates. During the cutting process the instrument should be held at a low rake angle to provide an adequate depth of cut. The study suggested that clinicians will require training before using the tools.

Cardoni [87] through a series of experimental investigations, using fresh bovine femur, demonstrated that the cutting temperature and hence thermal damage, can be reduced by selecting appropriate cutting parameters and ultrasonic blade profile. The responses were registered by six thermocouples positioned in the bone specimens, cut using 35 kHz blades with profiles 1 and 2 (Figure 3.19d) and with an applied load of 20 N. Significantly lower temperatures were recorded by all the probes when cutting with the blade with the indented profile (profile 2).
Figure 3.19  (a) Thermocouple locations in specimen; (b) blade profiles. Temperature responses measured in bone at six thermocouple locations using two 35 kHz blades with blade profile 1 and (c) blade profile 2 (d) [87]

The chisel-like action of an ultrasonic tool for the removal of bone was compared with a rotary burr and with a hand-held chisel by Horton [88]. The healing responses were found to be similar when using the chisel and the ultrasonic tool. The bone surface quality following use of the ultrasonic tool and chisel was rougher than that produced by the burr. The effects of an ultrasonic vibration, 28 kHz at 50 and 100 watts, imposed upon the longitudinal axis of a saw segment, were investigated by Thomas [89]. It was observed that a force reduction of up to 80 percent could be achieved over nonvibratory saw cuts.

Temperature measurements in a bone cutting process is of great importance since it determines the level of thermal necrosis and mechanical response at the implant-bone interface postoperatively. Various measurements techniques have previously been employed for this purpose and are discussed in detail in the following section.

3.7 Temperature Measurements in Bone Cutting

When cutting a solid material, energy is dissipated and heat is therefore produced. This heat is produced by the energy given off during irreversible deformation of the material and by friction between the material and the tool. Bone tissue’s low thermal conductivity and porous structure made it difficult for researchers to measure accurately temperature in
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the cutting zone. Temperature measurement near the bone region in contact with the cutting tools is widely reported in literature. The methods so far applied were bone-imbedded thermocouples, tool-imbedded thermocouples and infrared thermograph. Thermocouples have been used in the past in many experiments involving a temperature change. It is the most popular temperature measuring unit around, due to its relatively low cost and wide range of temperature sensing abilities.

3.7.1 Measurements with thermocouples

The use of thermocouples is widely reported in bone temperature measurements, especially in bone drilling. Attempts have been made to measure temperature when drilling bone by inserting thermocouples adjacent to the drill as it progressed into the hole. Krause [90] measured temperature when drilling bone by cementing a 1.07 mm diameter thermocouple into a hole previously drilled from the endosteal surface without irrigation. The drill under test approached the thermocouple along the drilling direction from the periosteum surface and stopped just short of it.

Thermocouples were cemented into four separate 0.7 mm diameter holes drilled about a centre at radii of 2.5, 3.0, 4.0 and 5.0 mm by Matthews [79][91]. A 4 mm hole was drilled through the centre and the temperature was measured with each thermocouple. Bachus [78] used three 0.27 mm (0.010 inch) diameter, Teflon insulated Type K thermocouples to measure fresh human cortical bone temperatures at locations 0.5 mm, 1.0 mm and 2.0 mm from the drilling tract. The tests were conducted in the presence of a supply of water at a temperature of 37°C. The experimental set up used in their experiment is shown in Figure 3.20.

**Figure 3.20** Experimental set-up of drilling human cortical bone [78]
Hillery [67] inserted a thermocouple into a drill bit and measured the drill temperature. A slip ring was used to counter the drill rotation. In order to place the thermocouple in the drill-bit as close to its cutting edge, a surgical stainless steel tube of 4.5 mm outside diameter having a bore of approximately 1 mm was used to make five drill-bits. A K-type insulated thermocouple was embedded inside the drill with its leads glued along the drill barrel. The wall thickness at the cutting edge was in the region 0.5–0.7 mm. In a recent study, Cardoni [87] utilised thermocouples in ultrasonic cutting of bovine femur. Temperature measurements were conducted using six thermocouples distributed in two rows of three, placed on opposite sides of the cutting line, as depicted in Figure 3.21. Thermocouples in each row were placed at distances of 5, 10 and 15 mm from the top surface of the specimen.

![Figure 3.21](image)

**Figure 3.21** Experimental arrangement for measuring cutting temperatures: (a) test rig for ultrasonic cutting experiments; (b) thermocouple locations in specimen [87]

### 3.7.2 Infrared thermography

Infrared thermography is a useful mean for recording temperature in circumstances where thermocouples or other contact type sensors cannot be used. A limited use of this technique is reported for the field of bone surgery/cutting. Benington [92] recorded thermal images using Thermovision 900 system for drilling bovine mandibular bone. Three drills used were round bur, spiral drill and pilot drill. Thermal pictures were recorded around the drilling areas. No irrigation was applied as infrared cannot pass through water. Denis [93] used an infrared thermometry in a robot assisted milling of human tibial bone for total knee arthroplasty. The thermal camera was focused on the freshly milled surface behind the end mill. The pyrometer was placed out of the way of bone particles.

The temperature distribution on the outer layer of the pig cortical bone in horizontal milling was calculated using thermography by Sugita [94]. The cutting temperature during the cutting process near the edge of the tool was measured. In the experiment, the cutter
proceeded in the radial direction with a prescribed depth of cut, and the cutting temperatures were measured until the cutting tool passed over the thermocouple and cut it off. Thermographic images of the temperature distribution at the cutting edge immediately after the end of machining by a single cutter blade are shown in Figure 3.22.

**Figure 3.22** Surface temperature observed by thermography in up and down cutting (cutting speed 31.4 m/min; down cutting; feed per tooth 0.375 mm/tooth; radial depth of cut 1 mm; axial depth of cut 5 mm; cutting tool – 2 flutes square end mill) [94]

### 3.8 Cutting Forces Measurements

Bone cutting forces in both rotational and linear tool moments were measured by load cells and load dynamometers. The forward and downward components of the applied forces were measured using a load cell and a strain gauge during bovine bone cutting with ultrasonic chisel by Khambay [9] (see Figure 3.17). In the experiments, a block of bovine bone was clamped on a metal beam with one arm attached to a Perspex stand. For the force to be measured in the longitudinal direction, a strain gauge was glued with epoxy resin to one side of the metal beam. The load cell and strain gauge were calibrated over a range of 0–600g.

Bachus [78] measured drilling forces in human cadaver femora using a 500 N load cell (A217-17A, Instron Corp, Canton, MA) anchored directly underneath the specimen. Four representative forces – 57, 83, 93 and 130 N were chosen for the study. Denis [93], while finding the influence of bone milling parameters on the temperature rise, milling forces and surface flatness, used a Kistler force platform to measure milling in robot-assisted total knee arthroplasty. An improved drill, which reduced the thrust force and allowed efficient removal of bone chips were reported by Saha [65] and Pal [95]. It was noted that drilling a predrilled hole significantly reduced the thrust forces due to the elimination of the chisel edge thrust at the tip of the drill.
In previous studies, the bone cutting processes have revealed critical variables and cutting parameters affecting mechanics of bone cutting. Jacob [60] showed that for all cutting directions relative to the bone matrix orientation, the cutting forces decrease with an increase in the tool rake angle, and with high rake angles the depth of cut had a negligible effect on the cutting forces. A similar relationship between the cutting forces and rake angle was reported by Wiggins [62]. The cutting forces were found to decrease with increasing rake angles and became increasingly nonlinear with larger depths of cut. The thrust force reached an asymptotic value using high rake angles for a depth of cut of 50 micrometers. For increasing rake angles, the specific cutting energy (total work required to cut the bone) was decreased with increasing depth of cut. Krause [96] expanded the conclusions reached for increasing the rake angle by also investigating the effect of cutting speed. It was reported that increasing the cutting speed decreased the cutting forces and the specific cutting energy. Therefore, the optimum cutting conditions would be: (1) a high rake angle; (2) a large depth of cut; and (3) a high cutting speed.

Forces in orthogonal cutting of bone were measured using tools of different materials and cutting conditions by Itoh [97]. One of the findings was that bone cutting processes differ from those of metal cutting due to the textile-like structure of the bone. In the case of human cortical bone, tools were made of quenched stainless steel, and the specific cutting force was found to be approx. 200 MPa. The cutting force was found to decrease with the increase in the tool rake angle except when the tool rake angle was –5°.

The relationship between cutting conditions and cutting parameters of pig cortical bone and a cancellous tissue, of which bone is comprised, were evaluated by Mitsuishi [98]. A bone cutting machine tool was developed suitable for minimally invasive surgery. The cutting processes on the bone were carefully observed and the specific cutting forces in radial, tangential and longitudinal directions against time were plotted. In that study, small tool rake angles were noticed to produced high cutting energy. Their experimental results are shown in Figure 3.23.
Figure 3.23 Cutting force measurements of pig cortical bone with endmill [98]

3.9 Cracking in Bone Cutting

Fracture mechanisms in cortical bone tissue under various loading conditions have been the subject of several studies [99][100]. Numerous techniques have been employed to monitor cracking phenomena in bone tissue including acoustic emission (AE) [101], sequential labelling [102], nonlinear resonant ultrasound spectroscopy (NRUS) [103], and micro-computed tomography [104] etc. In an effort to quantify microcrack growth and fracture in bone, indentation fracture has been used to measure fracture toughness at the microscale using scanning electron microscopy (SEM) [105] and optical microscopy [106]. While the fracture behaviour of bone has received considerable attention, analogous studies to crack monitoring using tool penetration are relatively novel.

The fracture of, and crack propagation in, the machining bovine cortical bone was observed and analysed using optical microscopy [107]. The cutting was implemented for three directions: parallel (along the bone’s major axis), cross and transverse. A crack was generated and propagated along the haversian canal while cutting was in the cross direction. Crack initiation was observed to be strongly affected by the relative position of the osteon and the cutting tool. Osteon were found to be compressed by the rake face of the cutting tool and produced a brittle crack at the layer boundary as shown in Figure 3.24(a). The crack was observed to propagate along the osteons as cutting continued. Generally, the crack was observed to emanate from the edge of the cutting tool and extended in a downward direction. In parallel direction, osteons and haversian canals were along the cutting direction. A crack propagated easily in parallel direction with the tool advance, and it removed a discrete layer of undamaged bone tissue, as shown in Figure 3.24(b). The crack direction was observed to be almost straight and was not changed in the cutting process. In the transverse direction, since osteons were perpendicular to the cut surface,
cracks elongated easily between the haversian canals and the boundaries of osteons, as shown in Figure 3.24(c).

**Figure 3.24** Optical micrograph showing crack propagation in each cutting direction: (a) across; (b) parallel; (c) transverse [107]

Indentation cutting was performed using sequential labelling and acoustic emission technique to analyse the damage in cortical bone during tool penetration [108]. Specimens of cortical bone cubes with a side of 8 mm were placed in a chamber and cut using a wedge-shaped blade in longitudinal and transverse directions (Figure 3.25). Chelating dyes were used at different loading stages. Specimens were sectioned and examined to visualise damage using UV microscopy post-test. The dyes label the progressive damage that occurred at various stages of the penetration process and thus indicated the extent of the damage zone in the bone around the cutting area. The observed damage zone was strongly influenced by the cutting directions; extending longitudinally for longitudinal cutting and confined to lateral regions for transverse cutting. For both cutting directions tested, fracture cracks was seen to grown along the lamella interfaces.

**Figure 3.25** Cutting zone – transverse cutting specimens [108]
4.1 Introduction

Finite element modelling is a main computational tool to simulate a cutting process and tool-work piece interaction in material cutting. It can provide detailed information on forces, stresses, temperature, chip shape and also predict an optimum set of parameters for an efficient cutting process. Finite element simulations allow the analysis of the cutting process in greater detail compared to experiments provided an adequate material data is available. Apart from forces, the computational cutting models can also predict shear angle, contact length and provide a platform for thermo-mechanical coupling for a tool-chip interface. Finite element codes can be used to simulate a very short process time before the onset of steady state conditions.

One of the state-of-the-art efforts in manufacturing engineering is the development of finite element simulations of the machining processes. Such computational models would have great value in advancing understanding of the cutting process and avoiding the number of expensive experiments, which are traditionally used for tool design, selection of optimum cutting parameters and related thermo-mechanical investigations. This chapter focuses on the basic elements of a finite element model of a cutting process, which can predict chip formation, cutting forces, temperature and pressure distribution on the tool-chip interface.

4.2 Eulerian vs. Lagrangian Formulation

The use of FE simulations has become increasingly popular due to the advancement in computers and the development of codes. In FE modelling, there are two types of analysis, in which a continuous medium can be described: Eulerian and Lagrangian [109]. In a Lagrangian analysis, the computational grid deforms with the material whereas in a Eulerian analysis, work piece material is assumed to flow through a meshed volume fixed in space [110]. Eulerian formulation offers the advantage of using fewer elements in modelling the chip formation, thereby reducing the analysis time. The Lagrangian formulation fixes a computational mesh to the material domain and solves for the position of the mesh at discrete points in time. This formulation requires a realistic separation criterion for simulating chip formation. Conversely, in an updated Lagrangian formulation, the need of such criteria is bypassed by using automatic remeshing schemes and has been successfully applied in simulation of chip formation in machining processes [111][112]. For problems involving large localized deformations and varying contact conditions as
those experienced in machining, a dynamic explicit FE formulation is very efficient for simulating the process.

The incompressible behaviour of fluids make Eulerian analysis a more preferred method in fluid mechanics than in solid mechanics [113]. It is recommended for processes where the shape of elements does not change. However, in a cutting process, the locations of drawing the elements is not clear and requires the development of free surface boundaries of the element mesh by iteration. The Lagrangian formulation is therefore a better choice for modelling a cutting process as it allows changes in the shape of the elements with material flow. The formulation requires updating of a strain matrix due to change in the elements shape leading to geometric non-linearities as well as material non-linearities.

4.3 FEM in Cutting Simulations

4.3.1 Chip formation

The chip formation process is similar for most machining processes. Modelling chip formation with finite element method (FEM) has proven useful for basic understanding and enhancing the machining processes. Coupled thermo-mechanical simulations of the chip formation process which predict cutting forces, stresses and temperature distribution in the chip, have been used by a number of researchers [111][114][115]. Simulations were mainly concerned with formation of continuous chips, however, some researchers have successfully applied FEM for modeling discontinuous/segmented chips [116][117].

Formation of continuous and segmented chips in machining alloy 718 was studied by Lorentzon [116]. The simulated chip shape with and without fracture criteria is shown in Figure 4.1. The effect of different fracture criteria on the chip formation process was

Figure 4.1 Chip formation process: (a) without fracture criteria - continuous chip; (b) plastic strain as fracture criterion – segmented chip [116]
studied, showing that the one with hydrostatic dependency should be used in simulations in order to capture the transition zone between continuous and segmented chips.

Modelling of discontinuous chip formation in hard machining was given by Guo [117]. A more general formulation – Arbitrary Lagrangian-Eulerian (ALE) – was used in simulations to maintain a high-quality mesh and to prevent analysis from divergence as a result of severe mesh distortion. The mechanism of discontinuous chip formation was simulated by creating an initial crack on the surface and extending it in front of the tool and then meeting with the surface crack. The study investigated that discontinuous chip can be efficiently simulated using the commercial FEA code. The cutting simulations showed the necessity of a damage model of the material as a mechanism for producing new free surface. The adaptive meshing along with the Johnson-Cook plasticity were found to be capable of crack initiation and propagation in the chip formation process.

4.3.2 Remeshing

In simulations of high-deformation problems, the mesh in the vicinity of the cutting zone becomes too distorted leading to the solution inaccuracy and premature termination of the analysis. It is therefore necessary to replace the distorted mesh with that of the correct shape. With Eulerian meshes, a pre-choice can be made regarding the area and level of refining the mesh and the intended sizes can be implemented easily. However, using a Lagrangian mesh for computing efficiency, there is a need to refine and then coarsen the mesh during the division of the elements as it flows into and out of high deformation zones with large shape distortions.

The automatic or manual procedures followed in remeshing are similar, with the process being divided into many loading steps. The mesh is checked after each loading step and all information regarding the nodes of the distorted elements in the meshed domain is stored in matrices. The distorted elements in the vicinity of and inside the high deformation zone are replaced by new elements with better shape. The stored nodal data from the previous step are mapped back onto the new nodes and the process continues. Most of the recent studies have utilised automatic remeshing and the results of cutting forces and temperatures were comparable to those obtained from experiments [111][114].

4.3.3 Modelling friction in cutting

In the analysis of a cutting process using FE simulations, predictions are greatly influenced by two major factors; (1) flow stress parameters of work piece material at cutting regimes and (2) friction characteristics, mainly at the tool-work piece interface
The difficulty arises in the implementation of accurate friction models for cutting simulations using a particular FE formulation. The exact definition of frictional conditions between the tool and the work piece is complex as well as sensitive to the cutting conditions. An accurate prediction of distributions of process variables such as forces, stresses and temperatures with FE simulations depends strongly on frictional conditions between the work piece and tool.

Thermomechanical analysis of orthogonal cutting was performed analytically by Moufki [118] with a friction law that accounted for temperature effects. The influence of cutting parameters and the material behaviour on the temperature distribution at tool-work piece interface were analysed. In that study the effect of mean friction $\mu$ on the cutting forces were also investigated. The Coulomb friction law at the tool-chip interface was considered with a friction coefficient as a decreasing function of temperature:

$$\mu = 1 - 3.441^{-1}T_{\text{int}} \quad \text{for} \quad 25^\circ\text{C} \leq T_{\text{int}} \leq 955^\circ\text{C},$$

$$\mu = 0.68 \left(1 - \frac{T_{\text{int}} - \bar{T}}{T_m - \bar{T}}\right) \quad \text{for} \quad 955^\circ\text{C} \leq T_{\text{int}} \leq 1500^\circ\text{C}$$

with $q = 1.7, \bar{T} = 955^\circ\text{C}, T_m = 1500^\circ\text{C}$ (melting temperature)

Experimental trends were reproduced in the analysis showing a decrease in the mean friction coefficient $\bar{\mu}$ with cutting velocity, the feeding and the increase of $\bar{\mu}$ in terms of the rake angle. The influence of friction conditions on thermomechanical quantities in a metal cutting process using numerical simulations was studied by Shi [119]. Modified Coulomb friction law was utilised in the model and the critical friction stress $\tau_c$ was calculated by

$$\tau_c = \min(\mu p, \tau_{\text{th}}),$$

Where $\mu$ is the coefficient of friction, $p$ is the normal pressure at the tool-chip interface, and $\tau_{\text{th}}$ is a threshold value for the conventional Coulomb friction stress. A value of friction coefficient ranging from 0.0 to 0.6 was used in the simulations. The curvature of the chip and the shear angle were decreased as the friction coefficient increased for a fixed value of the rake angle (Figure 4.2). The results of the simulations were consistent with those measured in experiments. Shear straining was found to be localized in the primary shear zone while the largest plastic strain rate was near the tool tip. The maximum
temperature rise and maximum effective plastic strain were found away from the cutting edge along the tool-chip contact interface.

Figure 4.2 Chip formation at time = 0.62 sec, for the case of rake angle $\alpha = 25^\circ$ and two values for the friction coefficient $\mu$ [119].

In the research by Özel [110], the influence of friction models on finite element simulations of machining low carbon free-cutting steel (LCFCS) was studied. The stress distributions on the tool rake face were experimentally measured and used in conjunction with metal cutting theory for the purpose of implementing different friction models. Five different friction conditions were used in simulations: (1) constant shear friction at the entire tool-chip interface (model I), (2) Coulomb friction in the sliding region and constant shear friction in the sticking region (model II), (3) variable shear friction at the tool-chip interface (model III), (4) a variable friction coefficient at the entire tool-chip interface (model IV) and, (5) a variable friction coefficient in the sliding region and variable shear friction in the sticking region (model V). The comparison of calculated process variables (cutting force $F_c$, thrust force $F_t$).

Table 4.1 Comparison of results of various friction models with experimental results for cutting speed of 150 m/min [110]
chip-tool contact length \( l_c \), shear angle \( \phi \) and maximum temperature at the tool-chip interface \( T_{\text{max}} \) from the process simulations using five models are presented in Table 4.1. The results showed that the chip geometry, forces, stresses on the tool and the temperatures at the cutting zone were significantly affected by various tool-chip interface friction models.

The level of the cutting force in any cutting operations is strongly influenced by the type of lubricant applied. While investigating the effect of lubrication on cutting force and temperature in ultrasonically-assisted turning of Inconel 718, a 3D thermomechanically coupled FE model was produced [114]. The objective of the study was to measure the cutting force for ideal lubrication conditions (minimum friction coefficient) at the tool-work interface. The calculated cutting force was 20–25\% lower in the absence of the friction force at the tool-chip interface. The temperature in the cutting region was by some 60\% lower due to the elimination of frictional heating.

### 4.3.4 Stress distribution in the chip

A coupled thermo-mechanical model of plane-strain orthogonal cutting of mild steel (0.18\% C) with continuous chip formation was produced to predict the stress distribution in the chip [120]. Contours of equivalent stress predicted during steady state cutting conditions are shown in Figure 4.3. A maximum value of stress was calculated at the secondary deformation zone, near to the sticking region (point 9 indicated by an arrow). The stress in the material was predicted to decrease gradually under the uncut surface.

![Figure 4.3 Contours of the equivalent stress at steady state in simulated orthogonal machining [120]](image)

### 4.3.5 Thermomechanical coupling
The history of the development of coupled thermo-mechanical FE models of cutting processes with inelastic material behaviour and a non-steady chip formation analysis goes back to the beginning of 1990s. Thermomechanical coupling is an essential feature of the numerical simulations of the machining process. During a cutting process, the temperature of the tool and material being cut significantly increases due to the work of irreversible deformation of material and friction between the cutting tool and work piece. As a result of temperature increase, the thermal and mechanical material properties, such as the yield strength, coefficient of thermal expansion, specific heat and conductivity are noticeably affected and are attributed to the change in microstructure of the material. These changes in the properties in turn will influence the cutting process in terms of stress evolution and temperature distribution in the work piece. Higher temperatures can lead to quick and excessive wear of the cutting edges that add additional cost to the manufacturing processes involved cutting. In studies devoted to numerical simulations of cutting processes, thermal coupling was usually achieved by implementing a staggered procedure where the analysis is divided into two steps i.e. thermal and mechanical. In the first step, thermal analysis is performed and the nodal temperature are calculated. In the second step, heat generated by plastic deformation and friction at the tool-work piece interface is used and the corresponding temperatures are recomputed, thus completing one cycle of the procedure.

A thermo-mechanical model of orthogonal cutting of 1020 steel was developed by Lei [121]. Temperature-dependent elastic-viscoplastic material properties with isotropic strain hardening were considered in simulation. The numerical domain accounted for coupling between the temperature field and plastic deformation. The coupled temperature and stress fields in the work piece and the chip during the cutting process were obtained using a non-linear temperature-displacement procedure. The thermomechanics of the process was simulated using a steady-state, two-dimensional form of the energy equation:

\[
\begin{align*}
K \left( \frac{\partial^2 T}{\partial x^2} + \frac{\partial^2 T}{\partial y^2} \right) - \rho C_p \left( u \frac{\partial T}{\partial x} + v \frac{\partial T}{\partial y} \right) + \dot{q}^{pl} &= 0, \\
\end{align*}
\]

where \( K \) is the thermal conductivity, \( h \) is the heat transfer coefficient, \( \rho \) is the density, \( u \) is the velocity in X direction, \( v \) is the velocity in Y direction, \( C_p \) is the specific heat and \( \dot{q}^{pl} \) is the volumetric rate of heat generation during the cutting process. The backward difference scheme was used for integration of temperatures, and the well established Newton’s method was utilised to solve the coupled system. The numerical model also accounted for non-linearities in the material behaviour, the geometry of elements and the
variations in contact conditions. For the solution to converge, the increments were automatically adjusted between minimum and maximum values.

An FE-based computational model was developed to determine temperature distribution near the cutting zone in the metal cutting process by Majumdar [122]. The heat generation in the deformation zones and at the tool-work piece interface caused by friction were incorporated into the FE model. Multi-dimensional steady-state heat diffusion equation in conjunction with heat losses by convection form the work piece was used in the numerical model. The heat generation was assumed to be uniformly distributed over the primary and secondary zone with convection losses approximated by constant convection transfer coefficients. The mathematical statement of the heat conduction process was given as

\[
K \left( \frac{\partial^2 T}{\partial x^2} + \frac{\partial^2 T}{\partial y^2} \right) + Q(x, y) = \rho C_p \left( u \frac{\partial T}{\partial x} + v \frac{\partial T}{\partial y} \right) = 0,
\]

which was subjected to the following boundary conditions:

\[ T = T_\infty, \]
\[ -K \frac{\partial T}{\partial n} = h_c (T - T_\infty) \]

Standard correlations for free convection and boiling for horizontal, vertical and inclined surfaces was used to obtain the convection film coefficients for the different surfaces involved. The heat generation and different convective conditions used on different surfaces of the model is shown in Figure 4.4.

**Figure 4.4** Heat source and boundary conditions in [122]
A thermomechanically coupled FE model of orthogonal cutting of mild steel with continuous chip formation was developed using finite element code MARC by Mamalis [120]. Figure 4.5 shows the temperature distribution in the work piece, chip and tool after a tool path of 2.58 mm. The highest temperatures in the chip were calculated in the secondary deformation zone (the sliding region). The temperature rise in the chip was largely attributed to plastic work that converted into heat. The highest plastic strains were also found at the secondary deformation zone.

![Figure 4.5](image1.png)

**Figure 4.5** Contours of temperature after a tool path of 2.58 mm in simulated orthogonal machining [120]

### 4.4 Material Separation Techniques

To simulate material cutting and chip formation, it is essential to introduce an adequate criterion of separation ahead of the tool tip, upon which the chip is developed. An appropriate selection of the criterion relies on the machining parameters and work piece’s material properties [115]. The choice of a separation criterion has been reported to have a strong influence on the deformation state and the residual stresses in the machined surface [123][124]. Various techniques have been employed to model the separation of material in cutting. Used approaches are: (1) defining a separation line and separating the nodes on this line when a certain criterion is met, and (2) a physical criteria which can use either a damage model or critical stress, critical strain to determine the separation [125]. The latter approach is closer to physical reality but complicated to implement in numerical modelling. In another approach no true material separation occurs and the cutting process is considered as a pure deformation process. Forging is an example of pure deformation
process. This approach allows the simulations to converge more easily as the problem of discontinuities is avoided. The “unphysical” phenomena associated with this technique is that the elements removed during remeshing steps can carry the load between the tool and the work piece [115].

4.4.1 Node separation technique

This technique has been used by many authors [123][124]. In those simulations, the nodes along the separation line were considered to initially have the same degrees of freedom. If a certain predefined criterion was met, the nodes were allowed to separate. Two-dimensional quasi-static finite element modelling of chip formation and shear banding in orthogonal metal cutting was studied in [123] with the FEA code, ABAQUS. The updated Lagrangian formulation for plane-strain conditions was used in that investigation. The chip was separated from the work piece surface using the ‘tiebreak’ slideline; The effective plastic strain was used as the material’s failure criterion.

Chip separation from the work piece was also simulated using material damage laws [126]. The Johnson and Cook damage law, which takes into account strain, strain-rate, temperature, and pressure, was used for 2D and 3D machining of 42CrMo4 steel using ABAQUS. The damage was calculated for each element and was defined as

$$ D = \sum \frac{\Delta \varepsilon_p}{\varepsilon_{pf}}, $$

where $\Delta \varepsilon_p$ is the increment of equivalent plastic strain during an integration step, and $\varepsilon_{pf}$ is the equivalent strain to fracture, under the current conditions. Separation was then allowed to occur when $D = 1$ and the related elements were removed from the analysis.

The general expression used for fracture strain was

$$ \varepsilon_{pf} = (D_1 + D_2 \exp D_3 \sigma^*)(1 + D_4 \ln \frac{\varepsilon_p}{\varepsilon_0}) \left[1 - D_5 \left(\frac{T - T_0}{T_{melt} - T_0}\right)^m\right]. $$

The constants of the JC fracture criterion $D_1 = 1.5$, $D_2 = 3.44$ and $D_3 = -2.12$, $D_4 = 0.002$, $D_5 = 0.1$ were identified from tensile tests.

A Similar approach was adopted in simulating cutting of aluminium alloy Al2024-T3 to study the nature of the stress state ahead of the tool [127]. Sacrificial elements were introduced in the model (Figure 4.6) along a straight narrow line that separated the area of chip formation from the work region. A ductile fracture criterion was used for failure of the
element. The JC damage parameters used in the simulations were $D_1 = 0.13$, $D_2 = 0.13$ and $D_3 = -1.5$, $D_4 = 0.011$, $D_5 = 0$.

**Figure 4.6** Mesh used in finite element analysis with sacrificial layer along [127]

### 4.4.2 Pure deformation technique

Material separation can also be considered as a deformation process. In this method, excessive deformations occur in the elements of the work material and the associated nodes move on the surface of the tool as it advances. The material that overlaps with the tool can be removed during a remeshing step so that the method also contains aspects of the element removal technique, but with sufficiently frequent remeshing it is possible to ensure that the amount of material removed remains small [115]. In that study, from the comparison of simulations with different remeshing frequencies, small variation was found.

**Figure 4.7** Equivalent plastic strains in two simulations performed with two different separation techniques showing a: (a) node separation method; (b) pure deformation method. The angle of shear band with horizontal is relatively larger in node separation method [115]
in the results. A comparison of chip shapes obtained with pure deformation and node separating techniques is given in Figure 4.7. The overall deformation of the chips and details of the deformation patterns for both simulations were found to be in good agreement.

Bäker [128] produced a finite element model of a two-dimensional, orthogonal metal-cutting process to study the influence of the cutting speed on the cutting force and the chip formation process with no frictional effects between the tool and the work piece. Material separation in the chip formation process was modelled using pure deformation technique with the material flowing visco-plastically around the tool tip. A small overlap of the elements of the work piece and the cutting tool in the cutting region was observed during the tool advance. The pure deformation technique was recently utilised in conventional and ultrasonically-assisted turning of Inconel 718 [129][130]. The level of cutting forces and temperatures predictions at the cutting zone matched the experimental results. Details of the literature describing material separation techniques for a number of materials and machining operations is provided in Table 4.2

<table>
<thead>
<tr>
<th>Reference</th>
<th>Cutting Type/Material</th>
<th>Separation Technique</th>
</tr>
</thead>
<tbody>
<tr>
<td>[8]</td>
<td>Ultrasonic cutting/foam and epoxy</td>
<td>Nodel debonding</td>
</tr>
<tr>
<td>[127]</td>
<td>Orthogonal cutting/Al2024-T3</td>
<td>JC damage model</td>
</tr>
<tr>
<td>[128]</td>
<td>Orthogonal cutting/Ti6Al4V</td>
<td>Pure deformation</td>
</tr>
<tr>
<td>[130]</td>
<td>Ultrasonic turning/Inconel 718</td>
<td>Pure deformation</td>
</tr>
<tr>
<td>[131]</td>
<td>Drilling/AISI 1020</td>
<td>Cohesive elements</td>
</tr>
<tr>
<td>[132]</td>
<td>Orthogonal cutting/Al 7075-T651</td>
<td>Strain rate threshold</td>
</tr>
<tr>
<td>[133]</td>
<td>Orthogonal cutting/bovine bone</td>
<td>Critical stress and COD</td>
</tr>
<tr>
<td>[134]</td>
<td>Orthogonal cutting/composite</td>
<td>Element deletion</td>
</tr>
<tr>
<td>[135]</td>
<td>Orthogonal cutting/FRP composites</td>
<td>Nodel debonding</td>
</tr>
</tbody>
</table>

4.5 UAC Simulations

All the research work presented in the previous sections was related to the cutting processes with non vibrating tools (conventional cutting). In the case of ultrasonic assisted cutting (UAC), the numerical model will require additional boundary conditions to
simulate the motions of vibrating tool. The section presents a review of the FE modelling of UAC.

The first successful attempt to model ultrasonically-assisted turning process of Inconel 718 using FEA, was presented by Mitrofanov [112]. The 2D model was developed using general FE code MSC. MARC and the material considered was Inconel 718 with elastic-plastic behaviour. Transient analysis was performed to study thermo-mechanics of the cutting process with constant cutting speed of 300 mm/s. Kinematic boundary conditions were applied to the left, right and bottom side of the work piece (AFGH in Figure 4.8) to simulate the vibration of the cutting tool. To simulate the motions of the tool in conventional cutting (CC), the cutting tool was assumed as rigid and fixed in space in numerical model. The vibration of the tool was considered in the direction of cutting in simulations of UAC:

\[ u_x = -a \cos \omega t, \quad u_y = 0, \]

where \( \omega = 2\pi f \), the frequency \( f = 20 \text{ kHz} \) and amplitude \( a = 13 \text{ \mu m} \).

**Figure 4.8** Scheme of relative movements of work piece and cutting tool in orthogonal ultrasonically-assisted turning [112]

Various contact conditions between the tool and chip in conventional and ultrasonic cutting and the friction type influenced the evolution of mechanical and thermal processes, causing difference in chip formation.

Later by Mitrofanov [129], A 2D thermo-mechanically coupled FE model of ultrasonically-assisted turning (UAC) was developed to simulate temperatures in the cutting regime (Figure 4.9). The model was incorporated with temperature-dependent material properties and strain-rate effects. The heat generation considered in the model was mainly due to the plastic work and friction at the tool/work piece interface. The results of
Figure 4.9 Temperature distribution in the cutting region in UAT ($h = 0.05 \text{ W/m}^2\text{K}$, $H = 500,000 \text{ W/m}^2\text{K}$, $\mu = 0.5$, $t = 2.9 \text{ ms}$) [129]

Simulations showed that the temperature in the cutting zone in conventional turning (CT) was about 15% lower than those in UAT. Temperatures were also measured in both types of cutting using thermography and the obtained results were found to be in good agreement with those predicted in simulations. It was shown that an increase in contact heat conduction between the chip and cutting tool significantly reduced the temperature of the work piece.

The authors studied turning of Inconnel 738 with a tool vibrating at ultrasonic frequency using FE simulations by Amini [136]. The study was concerned with the calculations of cutting forces and stresses in the material during the process. The influence of process parameters such as tool geometry, cutting speed and amplitude of the ultrasonic vibration on the level of force on the tool and stresses in the work piece was investigated. Periodic changes in the magnitude of the stresses and cutting force was calculated during simulations of UAT. The results indicated that the forces on the tool and the stresses in the work piece were inversely correlated with ultrasonic vibration amplitude and the cutting speed. The stress and force levels predicted for a cycle of ultrasonic vibration are shown in Figure 4.10. The tools with smaller rake angle ($10^\circ$, $15^\circ$ and $20^\circ$) resulted in higher cutting force whereas, clearance angle ($2^\circ$, $4^\circ$, $6^\circ$, $8^\circ$ and $10^\circ$) was found to have no significant effect on the magnitude of the cutting force. In that study, modal analysis was performed to get optimum configuration of the vibratory tool to deliver the required frequency for cutting. It was concluded that conical-cylindrical aluminium horns (110 mm) with concentrically attached tool insert provided more satisfactory performance.
4.6 FE Study of Bone Cutting

Finite element method is extensively used in machining of metals, polymers, composites etc. in the last nearly four decades. Despite capabilities of the finite element method in modelling materials, it has few applications in the area of cutting of biomaterials. So far, limited finite element studies of bone cutting have been reported in the literature. The studies performed are mainly concerned with modelling of bone (cortical and trabecular) orthotropic/anisotropic material properties, bone damage, bone cracking, bone remodelling and stress and stress strain estimations under various loading conditions [18][28][30][41]. The material and structural complexity is the main difficulty preventing bone cutting from being modelled with FEA. At a micro level, bone is of composite nature and knowledge of the cracking phenomena during tool penetration is necessary for developing a micro-mechanical FE bone cutting model in future. As bone shows similar material behaviour as that of wood in cutting, FE wood cutting models have been studied.

4.6.1 Cortical bone cutting simulations

As mentioned above, studies of bone cutting simulations are limited. In study by McCrohan [133], an orthogonal cutting FE model of bovine bone was suggested by considering bone as a naturally occurring fibre reinforced composite (FRC). Critical Stress Criterion and Crack Opening Displacement (COD) methods for chip separation were adopted. The finite element model produced is shown in Figure 4.11.
The authors in Ref.[107] analysed the fracture of and crack propagation in a cortical bone ahead of a milling cutter using FE simulations. The vector plots representing the maximum shear stress and minimum principal stress in simulations of cutting in the across direction (perpendicular to osteons) were shown. The cutting conditions were as follows: the tool rake angle of $20^\circ$ and the tool feed angle of $15^\circ$ or $75^\circ$. It was noted that the area around the cutting edge was exposed to tensile stress field for a small feed angle of the tool, e.g., $15^\circ$. Results of simulations presumed that a crack will propagate along the direction of minimum principal stress. On the other hand, compression stress state was predicted in the upward direction to the tool rake face when large tool feed angle was used, e.g., $75^\circ$. Under those conditions, microcracks were found, and a crack was initiated along the direction of maximum shear stress.

### 4.6.2 Wood cutting

Mechanics of cutting wood across the grain was examined by Le-Ngoc [137]. The effects of cell collapse during the cutting process of wood at the cellular level was studied using FE simulations. The model successfully simulated the compression test across the grain and associated behaviours of the cell collapse. The numerical model was developed with commercial finite element program, EMRC-NISA. A two dimensional plane-strain finite element model of the slicing test is shown in Figure 4.12. The model was then used to simulate two orthogonal cutting processes – wedge cutting and chip-forming cutting. The results showed buckling of cell walls ahead of the tool during the cutting process. A localized tensile stress in the longitudinal direction of the cell near the tool’s tip was caused by the collapse of the supporting cell wall leading to microfibrils being pulled-out of out of the matrix. The study concluded that the FE model of wood cutting would fail to simulate the material response at cellular level if the wood was assumed as a homogeneous material.
The influence of a CrN coating and the various cutting parameters on the average heat flux in a cutting tool during wood peeling process was highlighted by Kusiak [138]. For this purpose a finite element model was considered for three distinct geometrical configurations. In the first configuration, the contact area on the clearance face was absent, and the total heat flux was applied to the cutting face. In the second configuration, 70% of the flux was applied to the cutting face and 30% on the clearance face. The height of the contact area on the clearance face was equal to 0.7 mm. In the third configuration, 50% of flux was applied to the cutting face and 50% on the clearance face. The height of the contact area on the clearance face was equal to 1.4 mm. In three cases, the total heat flux was equal to 10 W. The study found the possibility to estimate very precisely the heat flux applied to the tool during machining when the contact area on the clearance zone evolves from 0 to 1.4 mm.

4.6.3 Composites cutting simulations

Numerous studies have been developed to investigate the failure mechanism and the level of cutting forces for FRC using FE simulations. A constitutive 2D force cutting model of a fibre-reinforced composite material was produced to investigate the variation of the cutting forces with fibre orientation [134]. Tsai-Hill criterion (for maximum work) was used to separate the material ahead of the cutting tool during the chip formation process. An equivalent homogeneous anisotropic material (EHAM) obtained from the properties of fibre and the matrix, was considered in the numerical model. The calculated forces were found to increase with fibre angle for 0° and 20° rake angle. Cutting forces were measured
with 4.5° rake angle under plane-stress conditions and were also studied for other fibre angles (orientation) as shown in Figure 4.13.

**Figure 4.13** Cutting force vs. fibre angle in anisotropic material [134]

Orthogonal machining of unidirectional carbon fibre reinforced polymer (UD-CFRP) and glass fibre reinforced polymer (UD-GFRP) composites was simulated using finite element method by Rao [135]. To predict the cutting force, a two-phase micro-mechanical FE model assuming fibre/s as elastic and the matrix as elasto-plastic properties was developed. Fibre failure was based on maximum principle stresses reaching the tensile strength. The forces were found to increase as the fibre orientation increased (from a parallel to a normal position) for both the composites as shown in Figure 4.14.

**Figure 4.14** Validation of cutting forces for UD-CFRP and UD-GFRP composites at $\gamma = 10^\circ$ and $t = 0.2$ mm [135]
4.7 Simulations of Drilling

Most of the studies have presented drilling models using either a mechanistic or a finite element approach. Despite complex kinematics of the drilling process, the finite element method has been successfully applied to model drilling of metals [131][139] and composites [140][141][142][143]. It can provide a unified approach for drilling with recent developments in FE codes enabling its 3D modelling.

4.7.1 Thrust force calculations

A number of studies reported the drilling thrust force and torque using finite element simulations [131][142][143]. Strenkowski [131] developed an analytical finite element method to predict the level of thrust force and torque in drilling AISI 1020 tube using a twist drill. The drilling was assumed as a series of cutting of oblique sections. Similarly, cutting of a chisel was considered as orthogonal cutting with different cutting speeds. For each section, a finite element model based on Eulerian formulation was used to simulate the cutting force of each section. The force produced by each section was added to calculate the overall thrust force and drilling torque.

A numerical FE analysis was proposed to calculate the magnitude of thrust forces in drilling of long-fibre composite structures those can produce defect at the location where the drill exit the hole [142]. The resulting thrust force \( F_z \) was broken up into two components \( F_{z1} \) and \( F_{z2} \) (Figure 4.15(b)), with \( F_{z1} \) as the thrust force induced by the contact of the cutting lips with the laminate and \( F_{z2} \) as the force corresponding to the contact of the chisel edge with the laminate (\( F_z = 2F_{z1} + F_{z2} \)).

Figure 4.15 (a) Decomposition of total thrust force \( F_z \), (b) cross view of mesh and boundary conditions. drill diameter 4.8 mm [142]
4.7.2 Temperature calculations

Calculations of drilling temperatures using finite element simulations have been the focus of several research studies [132][144][145]. The drill bit temperature of the AISI 1040 steel and Al 7075-T651 materials was analyzed using the finite element method in drilling [132]. The drill bit temperature was predicted with a numerical simulations using the ThirdWave AdvantEdge Lagrangian method based on the explicit finite element analysis software. The influence of the spindle speed and feed rate on the drill temperature was investigated; for linear changes of those parameters the temperature changed non-linearly. The drill heat flux, temperatures, and the thermal distortion of drill holes were predicted by Bono [144]. Heat flux was applied to the elements located under the cutting edge of the drill to represent heat flow into the work piece during drilling of aluminium 319. The diameter of the HSS drill considered was 9.92 mm with drilling speed ranging from 3000 to 7000 rpm and feeds ranging from 127 to 381 µm/rev. FEA was used to compute the temperature field and the thermally distorted shape of the work piece.

4.7.3 Drilling as 2D cutting process

The twist drill may be considered as a multi-cutting tool with two main and one secondary cutting edges. The cutting lip extends from the chisel edge and ends at the drill’s outer diameter. During the cutting, parameters such as rake angle, the inclination angle and the linear cutting speed vary along the drill radius. The drill bit geometry has a complex shape and the modelling of the cutting process is a complex matter. The known theoretical models are valid only for a small part of the cutting edges or the cutting being only on drill corner level [132][139]. The location on a drill-bit where the assumption of a two-dimensional cutting process can be applied (labelled A) is shown in Figure 4.16.

Figure 4.16 Cutting lips at the twist drill where orthogonal cutting model can be applied [139]
The drill bit’s temperature for AISI 1040 and Al 7075-T651 was analyzed using the finite element method by Ozcelik [132]. In simulations, the drilling process was assumed as a 2D orthogonal cutting at the outer edge of the drill cutting lip. In the analyses, the feed rate of 0.16 mm/rev with spindle speed of 955 rpm for AISI 1040 steel and feed rate of 0.2 mm/rev with spindle speed of 3184 rpm for Al 7075-T651 were used. Good agreement between the measured and analysed temperature results was found in dry drilling. The temperature data obtained from experimental study and FEM for AISI 1040 steel were observed as 350°C and 340°C and for Al 7075-T651 as 250°C and 255°C, respectively. The temperature distribution during the cutting process is shown in Figure 4.17.

Figure 4.17 Temperature distribution for drilling of AISI 1040 [132]

Another finite element model was developed to predict the temperature distribution in the drill, where the heat flux loads applied to the finite element model were determined from analytical equations [147]. To calculate the heat flux loads, the cutting action of the drill was assumed as a series of individual elementary cutting edges (Figure 4.18a). The

Figure 4.18 (a) Elementary cutting tools along cutting edges of drill; (b) predicted normalised temperature distributions in drill point (drilling depth of 25 mm) [147]
temperature profiles obtained were normalised with respect to the average temperature on the main cutting edges (Figure 4.18b). In the profile predicted, the maximum temperature was calculated on the chisel edge and the temperature was found to drop until the ends of the primary cutting edges were reached (Figure 4.18b).
Chapter 5. Characterization of Mechanical Properties of Bone

5.1 Introduction

The study of the mechanical properties of bone is important for the biomedical community, particularly to orthopaedic surgeons and biomedical engineers. One of the basic objectives of this study was to develop a FE bone cutting model capable of calculating the level of penetration force of the tool and temperature in the material. In addition, a response of the bone material to ultrasonic tool penetration is to be studied. To accomplish this, the respective input parameters such as material properties, frictional characteristics at the tool-work interface and an adequate material separation criteria are required.

As discussed earlier, that finite element modelling (FEM) of bone cutting could be a possible substitute for experimental work where expensive experimental equipments and health risk is involved. To our knowledge, limited attention has been given to the measurements of mechanical properties of bone to provide a primary input to a numerical model of bone cutting. Previous studies were largely focused on the mechanical and thermal properties of various types of cortical bones i.e. human and animals. Also to validate our FE model, the cortical bone properties used in the model and the data of cutting force and temperature measured in experiments were obtained using the cortical bone of the same animal and anatomical location (middle diaphysis of bovine femur). Our institution has initiated extensive studies to evaluate bone material properties for such modelling. Those include the response of the bone to various types of loadings including the investigation of fracture properties. Currently our FE model relies on the tension tests those measured the strain rate sensitivity in the bone material. The experimental programme to measure the mechanical properties of cortical bone tissue was implemented with the purpose of FE bone cutting model.

5.2 Uniaxial Tensile Test

Mechanical tests can provide a wide range of input for numerical models of bone mechanics, adaptation and repairs. Among the other tests, tensile testing is one of the most precise methods for measuring bone properties. These however, can be obtained provided the specimens are carefully machined from the region of interest. To measure strain rate sensitivity, the size of the specimens should be relatively large to enable sufficient grip in the holding devices since high loading rate can cause the slip of the specimen.
Machining/Cutting is a fast process and the strain rate in the material may reach up to $10^5 \text{ s}^{-1}$. Smaller specimens may not provide a firm grip in the holder of the equipment which can lead to slippage particularly in the case of high strain rate measurements. Uniaxial tensile tests are performed to obtain the stress-strain relationship as well as to study the strain rate effect on the pre yield and post yield behaviour of bone.

5.2.1 Experimental equipment

The required properties of bone tissue were obtained under tensile loading condition. Since the size of the bone specimens used in the tests were small, a high-precision tensile testing machine was required. Precise measurements with such specimens requires high precision equipments with a rigid frame that maintains alignment as well as high stiffness throughout the load range. Since standard universal testing machines are not recommended when small deformations are involved, the Instron MicroTester (5848) material testing system was chosen. It provides a solution to the challenges of testing microcomponent (Figure 5.1). MicroTester system combines a very stiff loading frame with displacement measurement at submicron accuracy to ensure performance [148].

![Figure 5.1](image)

The MicroTester, which is ideal for tests requiring less than 2 kN force, is offered with Instron's 5500 series controller and Bluehill-2 software for monotonic and simple cyclic applications. The equipment can be used with a wide range of grips and fixtures for various
tests, i.e. tension, compression and bending. For strain measurement accessories including extensometers are provided by the manufacturer. The equipment has two aligned columns fixed to the base and moveable crosshead. It contains a stiff reaction frame that can ensure precise deformation and an accurate displacement control. Instron provides a resolution of approximately 20 nanometers and displacement accuracy up to 0.5 micrometers. Instron provides a precise force measurement using load measurement transducers. The Instron 5800 series controller provides a load measurement accuracy of 0.5%. Bluehill-2 software used with the system is also interfaced with the data acquisition where an essential data such as displacement, load etc can be generated in ASCII format for further use. The experimental arrangement of the tension test is shown in Figure 5.2.

![Figure 5.2 Cortical bone tensile testing using MicroTester](image)

### 5.2.2 Specimen preparation

Tensile testing can be one of the accurate methods for measuring bone properties, but requires a relatively large and precisely machined specimens from the region of interest. The specimen shape most commonly used in tension test of cortical bone is so called dogbone type specimen although rectangular parallelepiped specimens have also been used [149].

The specimens used in our tests were excised from the middle diaphysis of the cortical bone with the specimen axis along the longitudinal axis of the bone using a disc cutter on
milling machine as shown in Figure 5.3(a). The rectangular cortical bone specimens were cut slightly larger than the final specimen’s dimensions to allow a clamping area for the milling process. The end area was sufficiently large to be gripped between the jaws of the holder and avoid slip. The test region or the gauge length was kept smaller in cross-section, the specimens should fail there. The final test specimen dimensions were based on both the previous literature and ASTM standards for tension and compression material testing [150]. A stream of cold water was employed during milling operation to avoid any thermal effects in the bone tissue. To produce fillets, end milling was performed. A total of 50 specimens were excised from 4 femurs of the same animal. After careful examination of the specimens, 20 specimens were rejected due to machining defects or visible porosity and the remaining 30 specimens were selected for the test. Finally, the specimens were evenly polished with 240, 320, 400 and 600 grit wet sand paper to produce a uniform cross section. Dimensions of the finished test specimen are shown in Figure 5.3(b). The end section are 8 mm wide, narrowing to 3 mm in the central waisted section, which is 10 mm long. Specimens were then kept refrigerated at -10°C until used in the experiments.

![Figure 5.3](image)

**Figure 5.3** Stages of specimen preparation for tension test: (a) rectangular piece excised from the cortical thickness along bone longitudinal axis; (b) specimen after machining to required dimensions. Thickness of the specimen is 2 mm. All dimensions are in mm

### 5.2.3 Experimental procedure

A tested specimen was firmly fixed between the upper and lower grips of the testing machine. In the experiments it was observed that specimens dried quickly after being taken out from a refrigerator. The specimens were kept wet by applying cold water during the test. An extensometer was attached to the specimen to record the level of strain at the gauge length (refer to Figure 5.2). To study the strain-rate sensitivity, the specimens were tested under different displacement rates. The displacement rate was varied between 0.0001 mm/s to 10 mm/s. A total of 30 specimens was tested at the prescribed heads
displacement rates. Though all the specimens were waisted in the central section, fracture in some specimens accrued in the shoulder region close to the clamps and data resulting from such tests was discarded. Results only of those specimens were considered which broke at the middle of the gauge length.

5.2.4 Tension test results

An elastic-plastic behaviour of cortical bone was observed in uniaxial tension tests. A number of studies of both cortical and trabecular bone have shown inelastic strain similar to our results, see e.g. [20][30]. A typical stress-strain graph obtained from the experimental data for a strain rate of 0.001 $s^{-1}$ is shown in Figure 5.4. Stress was calculated as the applied force divided by the cross section at the gauge length. The elastic portion of the stress-strain is a straight line (Hook’s law) and the slope of the line defines the elastic modulus. As the stress was increased the relation between stress and strain was no more linear showing plastic deformation in the bone. In tension, it yielded, followed by (linear) hardening up to an average failure strain of order 2.4%. Since the stress-strain curves were closed to each other for each strain rate, therefore, the variation in the stress is shown only on the onset of yielding and at failure strain. The failure of the specimen was observed primarily due to shear as shown in Figure 5.4(b).

![Figure 5.4](image)

**Figure 5.4** (a) Stress strain relationship of bovine cortical bone obtained from tension tests (b) shear failure of the specimen at the middle section. ($n = 10$)

**Strain rate sensitivity**

To determine a strain-rate dependency of the bone material, thirty specimens were tested for three strain rates between 0.00001 $s^{-1}$, 0.001 $s^{-1}$ and 1 $s^{-1}$ with 10 samples for
each strain rate. The specimens were not tested for strain rates higher than 1 s\(^{-1}\) due to slippage of specimen ends in the holding device. The strain rate was observed to have influence on the elastic modulus as shown in Figure 5.5. At a lower strain rate (0.00001 s\(^{-1}\)), the elastic modulus was measured to be 12 GPa. At the strain rates 0.001 s\(^{-1}\) and above the elastic modulus was approx. 21 GPa. The range of elastic modulus obtained was comparable with the results obtained from nanoindentation tests (17–20 GPa) described in section 5.3.5.

The yield stress was increased by 72% – from 58 MPa and 101 MPa – as the strain rate was changed from 0.00001 s\(^{-1}\) to 0.001 s\(^{-1}\) (Figure 5.5). The additional increase for the change in strain rate from 0.001 s\(^{-1}\) to 1 s\(^{-1}\) was below 7%. A similar behaviour of cortical bone of human and bovine femurs was reported in Refs.[23][24]. Based on results of those experiments it was assumed that the effect of strain rates higher than 1 s\(^{-1}\) can be neglected. A difference in fracture stress of about 62% was found between the strain rates 0.00001 s\(^{-1}\) and 0.001 s\(^{-1}\). The difference drops to 11% when the strain rate was changed from 0.001 s\(^{-1}\) to 1 s\(^{-1}\).

![Figure 5.5](image)  
**Figure 5.5** Elastic-plastic material behaviour of cortical bone with strain-rate dependency

From the obtained experimental results it can be concluded that the change in the elastic modulus, fracture stress and strain at higher strain rates can be neglected. This phenomena provided the opportunity to implement the stress-strain behaviour in bone cutting modelling/simulations, which may require high strain rates up to 10\(^5\) s\(^{-1}\). The magnitude of measured elastic moduls was in good agreement with those reported before. To obtain the effective modulus for our homogeneous FE model, nanoindentation tests
were carried out to see the difference in the elastic modulus at microstructure level (osteonal and interstitial matrix) and is described in the following section.

5.3 Nanoindentation

Understanding the mechanical properties of bone material in relation to its hierarchical structural organization is a complex problem to resolve. Nanoindentation is an advanced technique compared to conventional hardness tests for the assessment of the mechanical properties of thin films, layers, small volumes, and small microstructural features. Properties measured by nanoindentation could provide useful data in the development of theoretical micromechanical models, and finite element modelling. Since many of the microstructural components of cortical bone are several micrometres or more in size, the nanoindentation technique can be used to measure their intrinsic mechanical properties. Nanoindentation tests were performed to measure the elastic modulus and hardness of the cortical bone tissue and described in the next section.

5.3.1 Nano Test 600 indentation system

The NanoTest 600 system, manufactured by Micro Materials Wrexham UK, was used for the depth-sensing indentation (DSI) experiments. Figure 5.6 shows a schematic view of the NanoTest 600 unit used in the present experiments. The sample stage can be manipulated using precise DC motors having a displacement resolution of 17.3 nm in the X, Y and Z directions. The NanoTest system is supplied with a high-resolution microscope, with a microscope monitor, which helps to define the exact positions of the indents in a sample. The high resolution microscope has the ability to magnify in the range of 4X, 10X, 20X and 40X. Before starting the test, the “focal plane” and “measurement planes” needed to be calibrated. The “focal plane” was the plane parallel to the Y-Z plane of the sample holder, in which the surface of the sample was in focus at the highest magnification. Whenever the sample was changed, the sample was brought to the focal plane before starting an indentation test. The “measurement plane” was the plane parallel to the Y-Z plane of the sample holder in which the surface of the sample was approximately 50 µm away from the tip of the indenter.

A schematic of Nano test 600 indentation system is shown in Figure 5.6. The experimental equipment used in our study of bone samples is shown in Figure 5.7. The Nano head consists of a pendulum with an indenter having a load range of 0.1–500 mN and resolution of 0.1 µN. The pendulum can rotate on a frictionless pivot and is designed to be lightweight. The solid shaft of the pendulum is made of a ceramic, stiff enough to withstand the maximum load (500 mN) with negligible deflection. A coil is mounted at the
top of the pendulum and when the coil current is applied, the coil is attracted towards a permanent magnet, producing motion of the indenter towards the sample and into the sample surface. The parallel plate capacitor measures the displacement of the indenter as one of the plates is attached to the indenter holder. Thus, when load was applied and as the indenter displaced into the sample, the capacitance changed and the displacement was measured by means of a capacitance bridge. To minimise or reduce unwanted capacitance effects, the capacitance bridge unit is located close to the measuring capacitor. Below the capacitance plates, there is a counter-balance weight necessary to counter the mass of the coil and the indenter that is inserted into the diamond holder.

Figure 5.6 Schematic of Nano test 600 indentation system [151]

Figure 5.7 Experimental arrangement of bone nanoindentation test
The limit stop defines the maximum outward movement of the diamond as well as the operating orientation of the pendulum, when a load is applied. Its position is manually adjusted with a micrometer. The equilibrium position of the pendulum, with a zero load current, is adjusted with balance weights, movable along both the horizontal and vertical axes. DSI experiments can be performed by changing any of the following parameters: load, loading rate, unloading rate, dwell period – while keeping others constant.

5.3.2 Indenter type

Depth sensing indentation testing is generally made using spherical, conical or pyramidal indenters. In the present study, the Berkovich indenter was utilised as it has extensively been used for nanoindentation tests of cortical bone [36][40] (Figure 5.8). The face angle of this indenter at its apex is $65.2^\circ$ between the normal to the sides and $76^\circ54'$ between the normal to the edges which gives the same projected area to depth ratio as the Vickers indenter. The edges of the Berkovich indenters are easily constructed to meet at a single point and therefore are preferred over Vickers indenter [152].

Figure 5.8 Parameters of Berkovich indenters [152]

5.3.3 Specimen preparation

Parts of a cortical bone from the mid-diaphysis were cut down to 3 mm thick specimens with a low-speed diamond disc (1000 rpm) under constant water irrigation to minimise the undesired mineral formation on the surface of the specimen. The specimen was decalcified with HCl for two days to dissolve the minerals and to see the microstructure under microscopic observation. The specimen was then imbedded in Polymethyl methacrylate (PMMA) at room temperature as shown in Figure 5.9. The embedded specimen was metallographically polished using silicon carbide abrasive papers
with a decreasing grit size (600, 800, and 1200 grit) under de-ionised water to produce the smooth surfaces needed for nanoindentation testing. The specimens were polished on microcloths with successively finer grades of alumina powder of 0.05 mm grit, after which the specimen was ultrasonically cleaned to remove surface debris.

**Figure 5.9** Specimen used in nanoindentation tests: bone sample mounted into PMMA

### 5.3.4 Experimental procedure

The sample was mounted on the sample stage, having its surface parallel and vertical to the microscope and the nano head. All the tests on bone microstructure were performed to measure the elastic modulus and hardness at room temperature (23°C) and in the absence of an irrigation system. There were certain calibration and maintenance checks that needed to be performed for smooth functioning of the DSI equipment. The sample stage and zero-load calibration were performed before the experiments. The zero-load calibration did not require a sample to be present and was done automatically before each experiment in a schedule. The indentation parameters used were consistent with those reported before for cortical bone tissue [36][40]. The indentation depth and rate were 500 nm and 10 nm/s, respectively. Experiments were performed at a constant loading and unloading rate of 0.25 mN/sec to the maximum load of 12 mN, which produced impressions in bone with depths of about 1000 nm. The impression was held for a period of 1 minute at the peak load (dwell period). Our preliminary results showed that the elastic modulus was affected by less than 7% in the dwell period range from 10 seconds to 1 minute. The osteonal and interstitial bone locations were identified from microscopic images and the indenter was positioned on them. A total of 40 indents were made with 20 indents in osteon and interstitial matrix.

### 5.3.5 Nanoindentation test results

A typical force-displacement curve, which was used to calculate the elastic moduli and hardness of bone tissue, is shown in Figure 5.10. Those parameters were computed from
the unloading force-displacement curves with the Nanoindenter II software, according to the well established method described by Oliver [153].

![Figure 5.10 Force-displacement curve obtained by indentation of cortical bone](image)

The first step was to determine the contact stiffness, representing the resistance of the material to elastic deformation (see Figure 2.8, section 2.5.5):

\[
S(h_{\text{max}}) = \frac{dp}{dh}(h_{\text{max}}) = \frac{2}{\sqrt{\pi}} E \sqrt{A(h_{\text{max}})}
\]  

(5.1)

where \( p \) represents the applied load, \( S(h_{\text{max}}) \) is the derivative of the unloading curve at the point of initial unloading \( h_{\text{max}} \), which is determined by fitting 40–95% of the unloading curve, \( A, h \) is the contact area over which the indenter and the material are in instantaneous contact. The reduced modulus \( E_r \), depends on the deformation of the material and the diamond tip. It can be presented as follows:

\[
\frac{1}{E_r} = \frac{1 - \nu_{\text{bone}}^2}{E_{\text{bone}}} + \frac{1 - \nu_i^2}{E_i}
\]  

(5.2)

The definition of hardness was the mean pressure under the indenter at maximal depth. Calculation of the elastic moduli with Eq. (5.2) assumed a Berkovich diamond tip with an elastic modulus of 1141 GPa, a Poisson ratio of 0.07 and an isotropic, elasto-plastic bone sample property with a Poisson ratio of 0.3. The hardness of the material can be calculated from
The average measured elastic modulus and hardness of interstitial bone was larger than the osteonal bone respectively by 14% and 13%. The measured values of elastic modulus and hardness of bone microstructure are shown in Table 5.1. The average elastic modulus and hardness range obtained with nanoindentation tests compares favourably with those reported [36][37].

**Table 5.1** Elastic modulus and hardness measurements obtained from bone microstructure

<table>
<thead>
<tr>
<th>Test</th>
<th>Osteon</th>
<th>Interstitial bone</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Elastic Modulus (GPa)</td>
<td>Hardness (GPa)</td>
</tr>
<tr>
<td>1</td>
<td>14.4</td>
<td>0.31</td>
</tr>
<tr>
<td>2</td>
<td>11.3</td>
<td>0.72</td>
</tr>
<tr>
<td>3</td>
<td>17.1</td>
<td>0.55</td>
</tr>
<tr>
<td>4</td>
<td>21.2</td>
<td>0.43</td>
</tr>
<tr>
<td>5</td>
<td>16.5</td>
<td>0.49</td>
</tr>
<tr>
<td>6</td>
<td>16.3</td>
<td>0.62</td>
</tr>
<tr>
<td>7</td>
<td>15.8</td>
<td>0.59</td>
</tr>
<tr>
<td>8</td>
<td>16.7</td>
<td>0.51</td>
</tr>
<tr>
<td>9</td>
<td>20.4</td>
<td>0.63</td>
</tr>
<tr>
<td>10</td>
<td>20.2</td>
<td>0.36</td>
</tr>
<tr>
<td>11</td>
<td>17.7</td>
<td>0.44</td>
</tr>
<tr>
<td>12</td>
<td>18</td>
<td>0.56</td>
</tr>
<tr>
<td>13</td>
<td>16.5</td>
<td>0.63</td>
</tr>
<tr>
<td>14</td>
<td>18.2</td>
<td>0.43</td>
</tr>
<tr>
<td>15</td>
<td>15.8</td>
<td>0.36</td>
</tr>
<tr>
<td>16</td>
<td>16.9</td>
<td>0.62</td>
</tr>
<tr>
<td>17</td>
<td>19.4</td>
<td>0.55</td>
</tr>
<tr>
<td>18</td>
<td>20.2</td>
<td>0.51</td>
</tr>
<tr>
<td>19</td>
<td>19.2</td>
<td>0.53</td>
</tr>
<tr>
<td>20</td>
<td>16.7</td>
<td>0.36</td>
</tr>
<tr>
<td>Average</td>
<td>17.425</td>
<td>0.51</td>
</tr>
</tbody>
</table>

**5.4 Effective Elastic Properties**

Cortical bone is heterogeneous in nature with osteons embedded in the interstitial matrix and separated by a thin cement line. Topologically, it is similar to fibrous composites since osteons in it are aligned parallel to its axis. On the other hand, macroscopically bone is an orthotropic material with respect to the anatomical direction [25][28]. To obtain mechanical properties in each anatomical direction, particularly those related to higher strain-rates, is difficult since specimens only of shorter lengths (up to 10
mm in radial and transverse directions) can be machined from the bone. This makes the tension tests difficult to carry out for higher strain-rates in those directions. Therefore, the current model is relaying only on the tests performed in the longitudinal direction. The FE model assumes isotropy (both mechanically and thermally) of the bone material and that the cutting variables i.e. stresses, forces and temperatures are not significantly affected by the direction of cutting.

The increased complexity at the micro-level greatly complicates the analysis of the structural behaviour of a material such as bone. In the past several decades, numerous approaches have been proposed for the micromechanical analysis of composite materials (see [154] and references therein). These include the earliest rules-of-mixture approaches based on the Voigt and Reuss hypotheses. Much research has been focused on the mechanical behaviour of composite materials, and such studies often include theoretical or experimental determination of engineering constants. It is obvious that experimental tests and numerical models which often need significant effort and resources, can be avoided by the use of reliable rules.

Many authors have been trying to solve this problem by means of suggesting effective solutions using homogenisation techniques [155][156]. In fact, solutions can also be provided by the use of finite element calculations but these calculations are rather cumbersome [155]. Simple solution may be obtained by a more appropriate use of the classical rule of mixtures. The effective material properties obtained from the rule of mixture were assumed to be independent of the geometry, the boundary conditions and loading conditions of the macroscopic structure. In this way the effective material properties were assumed to be the intrinsic properties of the material when viewed macroscopically.

The rule of mixture is a mathematical expression, which gives some property of the composite in terms of the properties, quantity and arrangement of its constituents. The direct rule of mixture, named after Voigt, is based on the statement that the composite property is the sum of the properties of each constituent multiplied by the fraction of volume/area it occupies.

From the balance of forces:

\[ F_b = F_o + F_m = \sigma_o A_o + \sigma_m A_m , \]  

Where the subscript b, o and m represents the bone, osteons and interstitial matrix, respectively, and using the Hooke’s law follows:

\[ \sigma_b = E_b \varepsilon_b, \quad \sigma_o = E_o \varepsilon_o , \quad \sigma_m = E_m \varepsilon_m \]
$E_b = E_o A_o + E_m A_m \quad (5.5)$

Where $A_o$ and $A_m$ represent, respectively, the area fraction of osteons and interstitial matrix.

**Calculations for effective elastic modulus**

The concept of equivalent homogeneous materials has been successfully employed to evaluate the effective material properties of composites by Yang [157] and described in detail by Christensen [158]. In our study, a set of 10 microscopic unit cells of area each 1 mm\times1 mm were selected from the cross section of the prepared specimen to calculate the fraction area of constituents. i.e. osteons and interstitial matrix (Figure 5.11). The osteons were identified and manually encircled in the microscopic images. The images was then imported into Matlab programme where the area fraction of the constituents were measured. In this way the area fraction of each constituent in 10 cells were measured and averaged. The average area occupied by the osteons and interstitial matrix was 59\% (SD = 3.2) and 41\% (SD = 4.7), respectively. The differences in average measured elastic modulus and hardness of interstitial bone and the osteonal bone were 14\% and 13\% and provided a reason for idealising cortical bone as an equivalent homogeneous material. It is expected that these differences will not influence the cutting force acting on a cutting tool penetrating into the composite structure of the bone tissue. The effective elastic modulus was measured using equation 5.5 and was calculated to be 20 GPa.

![Figure 5.11 Measurement of elastic modulus from a unit cell: (a) microscopic image showing osteons embedded in interstitial matrix; (b) osteons are highlighted in composite structure; (c) black circles – osteons, white background – intestinal matrix](image)

**5.5 Concluding Remarks**

An experimental programme was performed to determine the mechanical properties of bones at macroscopic and microstructural level. Tension tests revealed the inelastic behaviour of bone consistentents confirming previous studies. The increase in the material’s
yield stress with the strain rate, will provide the opportunity to suggest an advanced material model that accounts for strain-rate sensitivity (such as the Jhonson-Cook material model, which includes the strain-rate effects). Results of nanoindentation tests showed the elastic modulus of interstitial matrix to be 17% higher than that of osteons. It was proposed to apply the classical rule of mixtures to provide a physically realistic representation of the bone’s composite structure. The rule of mixture with the assumption of isotropy helped in idealizing bone model as isotropic and an equivalent homogeneous material (EHM). The concept will be used in developing a FE bone cutting model, which will be presented in following chapters. Based on the obtained properties, the FE model is expected to resolve the questions related to the choice of parameters in orthopaedic bone cutting.
Chapter 6. Experimental Study of Conventional and Ultrasonically-Assisted Bone Cutting

6.1 Introduction

Bone cutting has always been a challenge to surgeons as it requires skills and precise instruments for successful implementation of operations. Mechanical operations performed on the bone tissue are similar to those on other materials, for example, high-speed burr cutting, drilling, reaming and sawing. Common complications include control of the cutting force and temperature during incision and were the main focus of the previous studies [78][79][87]. Force control enables smooth penetration of a surgical instrument to avoid the unnecessary damage to soft tissues surrounding the bone. Previous research carried out aimed at improving instrument design and controlling cutting parameters to reduce the risks associated with bone cutting. Recent technological advancements in ultrasonically-assisted cutting techniques has affected the design and development of optimal surgical instruments. Implementation of the cutting processes on bone has identified critical variables and cutting parameters, which affect the mechanics of bone cutting. Various factors that influence the cutting of bovine bone by an ultrasonic chisel were analysed, in-vitro by Khambay [9]. They concluded that when cutting bone with an ultrasonic chisel, both low forces and cutting rates were required.

Previous research conducted was mainly concerned with forces in bone drilling [79][86][95]; little attention has been paid to measure the level of force in a cutting process with a chisel used to score, cut, clean and sculpt the bone. The chisel does cut bone in a different way as compared to a conventional bur or drill, and the process should be made safe and efficient. The force transmitted to the bone is not always appropriate to generate the required cut using these tools [159]. There is an increasing concern to minimise the cutting force which may cause an injury to nerves in the area.

This chapter focuses on the level of force experienced by a tool cutting the bone in conventional and vibration modes. An experimental programme (with no irrigation environment) was performed to measure and compare the cutting force for two main types of cutting, i.e. plane cutting (chiselling action) and drilling. Another aim was to observe the effect of ultrasonic vibrations superimposed on the tool’s movements on the cutting process, particularly in drilling, which is the most frequent operation in orthopaedics. The cutting and ultrasonic vibration directions for both types of cutting and tools and modes are shown in Figure 6.1. The entire experimental work programme was carried out in
Loughborough University’s lab devoted to the development of ultrasonically-assisted machining technology.

![Figure 6.1](image1.png)

**Figure 6.1** Types of cortical bone cutting: (1) plane cutting; (2) drilling perpendicular to bone axis; (3) drilling parallel to bone axis. $X_1$ – radial direction, $X_2$ – transverse direction, $X_3$ – longitudinal direction (bone axis)

### 6.2 Construction of UAC System

A test rig for ultrasonically-assisted machining with autoresonant control system has been designed by Nonlinear Dynamics Group, Wolfson School of Mechanical and Manufacturing Engineering, Loughborough University, UK. The autoresonant system is a self oscillating positive feedback control system that maintains the drive to the transducer at its resonant frequency. The resonant frequency of the transducer depends on its length and on the velocity of sound in the material. The amplitude attained at the peak of resonance depends on the internal damping of a material. Aluminium is chosen as it offers less damping to the elastic waves. The system can be attached to a standard lathe or milling machine for various machining operations such as turning, drilling and plane cutting with a specially designed ultrasonic attachment. This prototype is used in all our experiments of bone cutting. A typical arrangement for applying ultrasonic vibration to the cutting tool is shown schematically in Figure 6.2.

The main elements of a UAT system are: (i) a high frequency generator, (ii) a transducer, which utilises the piezoelectric or magnetostrictive effect, (iii) a concentrator, which is shaped to amplify the vibration output of the transducer, (iv) a tool holder and (v) a tool. By definition, a transducer converts energy from one form to another. Transducers used in ultrasonic machining convert electrical energy into mechanical motion, and they are based on piezoelectric or magnetostrictive principles [160]. The ultrasonic transducer
contains two piezo-ceramic discs of PZT (Lead zirconate titanate) material. PZT is a ceramic material that provides a piezoelectric effect. The basic elements of which PZT-based material are composed are lead, zirconium and titanate combined at elevated temperatures. PZT-based compounds are also used in the manufacturing ceramic capacitors and STM/AFM actuators (tubes). The piezoelectric effect is produced by the voltage difference across two of the faces of the ceramic. PZT has an extremely large dielectric constant of magnitude >0.5 at the morphotropic phase boundary (MPB) making a unique electro-ceramic.

The amplitude of vibration depends on the length of the transducer and the strength of the material used. The practical limit of the transducer constructed was approximately 40 µm. The concentrator’s length is a multiple of half the wavelength of sound in the concentrator’s material. The reduction in the cross-sectional area causes amplification of vibration, which increases in inverse proportion to the ratio of the areas of the opposite ends of the concentrator. A sevenfold increase in vibration amplitude may be obtained with a suitably shaped concentrator [160]. The schematic of the components of the ultrasonic cutting system is shown in Figure 6.3.

Figure 6.2 Schematic diagram of ultrasonically-assisted cutting (UAC) system (after [160])
Figure 6.3 Components of ultrasonically-assisted cutting system

Figure 6.4 illustrates the UAC cutting mechanism for a system with a work piece cutting velocity $v_c$ and oscillating tool along the cutting direction at a frequency $f$ with an amplitude $a$ and the time period $T$. In UAC the tool-work piece contact is intermittent and remains for a certain period of vibration cycle. In Figure 6.4, it has been shown that the cutting occurs during the interaction periods $t_a$ and $t_b$ and is assumed constant for the duration of the contact time. The cutting force is produced only in this period of a tool vibration cycle.

Figure 6.4 Mechanism of the pulsating cutting force in ultrasonically-assisted cutting (UAC) [161]
The displacement \( x \) and tool vibration speed \( v_t \) are given by

\[
x = a \sin \omega t = a \sin 2\pi ft, \\
v_t = x = a \omega \cos \omega t.
\]

The vibrational cutting condition is satisfied if the tool speed is more than the work piece cutting velocity, i.e. \( 2\pi f a \pi > v_c \). Figure 6.4 also illustrates the following basic equations

\[
a \sin \omega (T + t_a) - a \sin \omega t_b = v_c (T + t_a - t_b). 
\]

Introducing \( r = \frac{t_c}{T} = \frac{t_b - t_a}{T} \); where \( r \) is the ratio of the cutting time to the cycle period, \( T \) and is known as contact ratio. These relations can be written as [161]

\[
\frac{af}{v_c} = \frac{1 - r}{2 \sin \pi r \cos \left( \cos^{-1} \left( \frac{v_c}{2\pi f} \right) - \pi r \right)} \quad \text{(for } 2\pi f > v_c \text{).} 
\]

Equation 6.3 shows that \( r \) during ultrasonic cutting is dependent on \( f, a \) and \( v_c \). As the work piece is cut for a certain period in each vibration cycle, the average cutting force also should be \( r \) times that of CC. This implies that the cutting force can be reduced exponentially by increasing ultrasonic frequency and amplitude. Experimental results obtained from conventional and vibration cutting are discussed below.

### 6.3 Plane Cutting Experiments

Plane cutting experiments similar to chiselling were conducted to investigate the effect of ultrasonic vibrations of the tool on the magnitude of cutting forces. The cutting tool was made of stainless steel as it is the base material for orthopaedic surgical tools. The tool width was selected as 3 mm and the nose radius as measured from microscopic observation was approximately 10 micrometers and wedge angle of 65°. Due to the shape and small surface cutting length (up to 80 mm) the maximum tool speed was such that it completed the travel in more than one second for the forces to stabilise and be measured.

#### 6.3.1 Specimen preparation

A fresh cortical (compact bone) sample was cut from bovine femur. The bovine bone was of interest since it replicated the properties of human bone according to [43]. The bone was obtained from a local butcher and was stored frozen at -10°C before the experiment. Epiphysis was then cut off with a hacksaw thus leaving bone diaphysis to be tested. The
average thickness of the cortical wall was 8–9 mm. The cylindrical columns of bone were approximately 80 mm in height and approximately 60 mm in diameter. The shape of the bone was not suitable to be gripped in a holding device on a lathe for plane cutting operation. To eliminate this problem, the bone was cut into two parts along its longitudinal axis. One portion of the bone was glued to the surface of a metal block with David Isopon P40 kit with the bone top surface facing the tool. The bone surface was usually not flat, hence a stable plane cutting operation across it was not possible due to the change in the DOC with the tool advance. This was resolved by planing the bone surface using a milling cutter as shown in Figure 6.5. The milling operation was also useful in removing the periosteum from the outer surface of the specimen as it was observed to cause the tool to block in our initial tests.

### 6.3.2 Experimental equipment

A conventional lathe machine (Harrison 300) was utilised in plane cutting experiments. An experimental setup for measuring the cutting force is shown in Figure 6.6. The metal block glued to the bone was fixed in a fixture and was mounted on a three-component force dynamometer. The depth of cut (DOC) was varied by changing the relative position of bone surface with regard to the tool. A cutting velocity was applied to the specimen against the tool. The transducer was fixed on the bed of the lathe by a specially designed tool post attachment, so that it was possible to apply ultrasonic vibration in the longitudinal direction. To keep recorded forces uninfluenced by machine vibrations, the natural frequency of the dynamometer should be kept large compared to the frequency of the exciting system [130]. For the force measurement, the quartz three-component
Dynamometer (Kistler Instruments AG, Type 9257A) was used in the experiments. The piezoelectric dynamometer was bolted at the bottom work piece holding fixture on the Harrison 300 lathe.

![Experimental setup for measuring cutting forces in plane cutting. Inset: enlarged cutting region](image)

**Figure 6.6** Experimental setup for measuring cutting forces in plane cutting. Inset: enlarged cutting region

There are two main ways of force measurements in metal cutting: direct and indirect [160]. In direct methods, electrical signals proportional to the cutting forces, deflections or strains are measured in elements supporting the cutting tool. These methods provide accurate measurements of forces both in magnitude and direction as compared to the indirect method where the lathe power is utilised. The description in this section is limited to the direct method. The dynamometer comprised four three-component force sensors, hence all three force components could be measured. The output voltage of the built-in charge amplifier is proportional to the applied force. A high rigidity of the system provides measurements practically without displacement. The dynamometer was calibrated with a spring balance in the quasistatic mode prior to dynamic force measurements in cutting. To keep recorded forces unaffected by machine vibrations, the natural frequency of the dynamometer should be kept large compared to the frequency of the exciting system [160]. The Picoscope series 2000 with a maximum frequency of 10 MHz was used for measuring the force in a digital format.

**6.3.3 Experimental conditions**

Experiments were carried out at room temperature (20°C) without additional cooling (irrigation). The cutting parameters and their magnitudes used in experiments are given in
Table 6.1. Cutting forces were measured and compared for both conventional plane cutting (CPC) and ultrasonically-assisted plane cutting (UAPC). The maximum cutting speed was kept at 16 mm/s, which was lower than usually used in surgery. This was the maximum speed that can be reached in the experiment to obtain the force data due to a short length of cut (about 80 mm) across the bone surface available for plane cutting. The cutting direction was along the longitudinal axis of the bone. The cutting force was measured with the force dynamometer placed beneath the fixture holding the specimen. The time taken by the force to stabilise (at the peak) was noted to vary with DOC. The downward cutting forces were negligible compared to those in longitudinal (forward) direction and were not analysed in this study. Various parameters, such as the DOC, cutting velocity, frequency and amplitude were changed to observe their effect on the level of cutting forces.

### Table 6.1 Parameters used in plane cutting experiments

<table>
<thead>
<tr>
<th>Parameters</th>
<th>CPC</th>
<th>UAPC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cutting speed (mm/s)</td>
<td>4 –16</td>
<td>4 –16</td>
</tr>
<tr>
<td>Tool rake angle (degrees)</td>
<td>20</td>
<td>20</td>
</tr>
<tr>
<td>Tool wedge angle (degrees)</td>
<td>65</td>
<td>65</td>
</tr>
<tr>
<td>DOC (mm)</td>
<td>0.1 – 0.3</td>
<td>0.1 – 0.3</td>
</tr>
<tr>
<td>Vibration amplitude (µm)</td>
<td>N/A</td>
<td>5 – 25</td>
</tr>
<tr>
<td>Vibration frequency (kHz)</td>
<td>N/A</td>
<td>10 – 30</td>
</tr>
</tbody>
</table>

6.3.4 Forces in plane cutting

Unlike numerical simulations (see Section 9.3.1), the force measurements were limited to three values of DOC as it was very difficult to control the exact depth within a narrow interval (0.1 mm and 0.3 mm) in real-life tests. In the tests, the cutting force increased linearly to a maximum value and became stable for the remaining length of cut. Continuous chips were produced in CPC up to a cutting depth of 0.3 mm and was no longer continuous when DOC was increased above 0.3 mm. Also using a DOC above 0.3 mm, the force was significantly large producing vibrations in the cutting equipment and was not used in the final experiments. Chip formation as a result of the flow process dependent upon the cutting direction and depth of cut was also observed in orthogonal cutting of bone by Jacobs [60].

In UAPC the observed chips were segmented for all values of DOC. A detailed analysis of the chip formation process in CD and UAD is discussed in Section 6.5.3. The force in both types of plane cutting (CPC and UAPC) was seen to oscillate at the peak as
shown in Figure 6.7. This may also be the result of vibration/dynamics in the cutting process as well as inherent errors of the measuring system. All the peak values in the subsequent plots are smoothed (averaged) for the steady portion of the graph.

![Force oscillation pattern at the peak in plane cutting obtained from dynamometer. DOC = 0.2 mm, cutting speed \( V_c = 10 \text{ mm/s} \)](image)

**Effect of DOC**

The experimental results showed a significant difference in forces acting on the cutting tool in CPC when DOC was varied. An increase of 104% was observed when the DOC was increased from 0.1 mm to 0.2 mm with the force growing by 69% when the DOC was changed from 0.2 mm to 0.3 mm. This can be explained by the fact that at larger DOCS the tool has to remove more material volume from the work piece. The increase in the frictional effect (due to a larger tool-work piece contact) is also a contributing factor. The magnitude of the longitudinal cutting force for different DOCs is shown in Figure 6.8.

In the next stage of experiments, vibrations were imposed on the cutting tool. UAPC showed a similar trend as observed in CPC when DOC was varied. An increase of 115% was observed when the DOC was increased from 0.1 mm to 0.2 mm. The increase of 90% was measured when it was changed from 0.2 mm to 0.3 mm. Cutting forces were decreased markedly when ultrasonic vibrations were imposed on the tool. The decrease in cutting forces for DOC = 0.1 mm, 0.2 mm and 0.3 mm were measured to be 45%, 42% and 34%, respectively. The effect of ultrasonic vibrations on cutting forces for different DOC is shown in Figure 6.8.
Figure 6.8 Comparison of plane cutting forces in CPC and UAPC for different DOCs. Frequency $f = 20$ kHz, amplitude $a = 10$ micrometers, cutting speed $V_c = 10$ mm/s

**Effect of tool velocity**

No significant change in the cutting force was observed in the experiments for the chosen range of velocities (Table 6.1). To account for the change in the cutting force with an increase in the material removal rate, the cutting velocity $V_c$ was varied from 4 mm/s to 16 mm/s. This was attributed to an unchanged friction coefficient at tool-bone interface, which affects cutting force, at lower cutting speed. Also no temperature rise was observed from the thermocouple inserted near the tool cutting edge which may be the reason for a condition of unchanged friction at the tool-bone interface.

**Effect of ultrasonic vibrations**

The effect of tool vibrations on the cutting force in UAPC was studied using the sets of variables provided in Table 6.1. To investigate the effect of ultrasonic frequency on cutting force, the DOC, amplitude and cutting velocity were kept constant. Here and below, each data point in the graphs represents the set of five experiments. The force was observed to drop linearly with increasing frequency and is shown, for three values of DOC, in Figure 6.9. The force dropped, respectively, by 81% and 38% using DOC 0.1 mm and 0.2 mm when frequency was increased from 10 to 30 kHz. The effect of frequency above 20 kHz on the force was negligible for larger DOC (0.3 mm), where for the case of smaller DOC, the drop was more significant.
Figure 6.9 Comparison of plane cutting forces in UAPC for three values of DOC (amplitude $a = 10$ micrometers, cutting speed $V_c = 16$ mm/s)

In UAC, the tool and the work piece kept separating and contacting producing the pulsating cutting force. The increasing ultrasonic frequency (for other cutting parameters being constant) decreased the contact ratio which produced a discretely pulsating cutting force. The pulsating force has been reported to promote the formation of cracks in the vicinity of the tool edge in metals [162]. The decrease in the contact ratio resulted in a drop in energy needed to remove materials that is caused by the impact force being applied over a shorter period of time. The decrease in contact ratio with increased vibration decreased the average value of force over the time of a force cycle. At higher DOCs, the scatter was prominent due to vibrations in the experimental setup.

An impact of the increasing amplitude on the cutting force in UAPC was also studied. Figure 6.10 presents the difference in forces resulting from changing the amplitude value from 5 µm to 25 µm for a constant frequency and cutting speed. An increase in the amplitude caused the decrease in the maximum force as shown in Figure 6.10. A significant drop in the magnitude of force was caused by the increase in the value of the amplitude from 5 µm to 15 µm for all values of DOC, while an amplitude above 15 µm (to 25 µm) did not decrease the cutting force. The decrease in cutting forces for DOC = 0.1 mm and 0.3 mm were measured to be 66% and 28%, respectively, using amplitudes 5 µm to 15 µm. The results revealed that the impact of amplitude on the cutting force was significant up to a certain level, and the force was not affected by a further increase in the amplitude for the range of DOC used.
After successful implementation of ultrasonically-assisted cutting technique for simplified plane cutting of bone, it was planned to further investigate its benefits in the most frequent surgical operation i.e. drilling.

6.4 Drilling Thrust Force Measurements

In this section, the drilling thrust force and torque in CD and UAD, for various drilling parameters and conditions are measured and compared.

6.4.1 Experimental equipment

The reaction force and torque generated during CD and UAD were studied using the experimental arrangement shown in Figure 6.11. The experiments were carried out on a three-axis CNC milling machine (Wadkin Machine Tools, UK). The maximum spindle speed of the machine is 10,000 rpm and feed 10 m/min. A two-component dynamometer (Kistler type 9271A), which can measure thrust force and torque, was used. The bone samples were glued to a metal plate and were clamped on the top of the dynamometer to provide rigidity. The force and torque signals generated by the Kistler dynamometer were conditioned using Kistler charge amplifiers and as in plane cutting tests, the signals were captured using a digital oscilloscope and transferred to the PC for subsequent processing.
6.4.2 Experimental procedure

All the experiments were performed at room temperature without additional cooling, and the drilling direction was perpendicular to the bone longitudinal and transverse axis (along the radial direction of the cortical core). Prior to the experiments, the dynamometer was calibrated and found to be accurate and linear in response to both forms of loading (torsional and longitudinal). Firstly, the drilling force and torque were measured under conventional conditions (no ultrasonic excitation) for a set of drilling parameters. Drilling tests were then performed with the ultrasonic transducer switched on for the same parameters used in CD. Many experiments were performed so that the data obtained could be averaged to give representative results. All the data points on the subsequent plots are the mean of five experiments. The drill bit was changed after drilling 20 holes to exclude the effect of wear that can influence the thrust force and torque. Drilling parameters used in the experiments are provided in Table 6.2.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>CD</th>
<th>UAD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Drilling speed, N (rpm)</td>
<td>600 – 3000</td>
<td>600 – 3000</td>
</tr>
<tr>
<td>Drill diameter, D (mm)</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>Drill cutting edge angle (degrees)</td>
<td>65</td>
<td>65</td>
</tr>
<tr>
<td>Feed rate, ( f_r ) (mm/min)</td>
<td>10 – 50</td>
<td>10 – 50</td>
</tr>
<tr>
<td>Vibration amplitude, ( a ) (µm)</td>
<td>N/A</td>
<td>5 – 25</td>
</tr>
<tr>
<td>Vibration frequency, ( f ) (kHz)</td>
<td>N/A</td>
<td>10 – 30</td>
</tr>
</tbody>
</table>

Table 6.2 Parameters used in bone drilling experiments
6.4.3 Results and discussion of drilling experiments

A typical force-time graph obtained with the measurement system is shown in Figure 6.12. After the initial engagement the force gradually increased with time and attained a plateau when the drill lip was fully engaged with the bone. Small oscillations were recorded at peak values due to the sensitivity of the measurement system and vibrations in the drilling equipment. Drilling thrust forces were measured and compared in CD and UAD. In this study, the drill vibration frequency was limited to 30 kHz since above this value overheating of the bone tissue occurred and fumes were produced. The effect of drilling speed on the thrust force was measured in CD and UAD and is shown in Figure 6.13. The thrust force decreased with the drilling speed for both drilling techniques (CD and UAD) for magnitudes above 1800 rpm. The decrease of the drilling force when the drilling speed was increased can be explained by the decay of the mean friction coefficient at the drill-bone interface. This may also be due to the chip formation mechanism at higher speeds that can affect the drilling thrust force. UAD resulted in a significant decrease in thrust forces for the range of speeds and feed rates used. The average value of the force measured was 66 N and 48 N for 600 rpm and 3000 rpm, respectively, in CD and was 36 N and 21 N in UAD for the same drilling speeds and feed rates. A decrease in force by 27% was observed when the drilling speed was varied from 600 rpm to 3000 rpm. With the vibrating drill, the force dropped by 45% for 600 rpm and 55% for 3000 rpm. Our results are consistent with those obtained by Hillery [67], where the thrust force was shown to drop exponentially with drilling speed in drilling bovine cortical bone (for a 3.2 mm drill). In that study the thrust force dropped from 48 N at 400 rpm to an asymptote of 23 N at 2000 rpm. The

![Figure 6.12](image)

**Figure 6.12** Force oscillation pattern obtained with dynamometer in drilling. Drilling speed $N=1800$ rpm; $f_r=40$ mm/min
exponential drop of force in our bone drilling experiments may be explained by the decay in the mean coefficient of friction at the drill-bone interface due to higher temperatures produced with higher drilling speeds (see section 7.3.4 and 7.4).

Figure 6.13 Influence of drilling speed on thrust force in CD and UAD, \( (f_c = 40, \text{ mm/min} \) frequency \( f = 20 \text{ kHz}, \text{amplitude } a = 10 \text{ micrometers}) \)

The influence of feed rate on the drilling thrust force was also studied. The force declined by a significant magnitude with the decreasing feed rate in CD as shown in Figure 6.14. Interestingly, but not surprisingly, the force level was not affected in the case of the ultrasonic vibrations for the range of feed rates used. The reason was the large difference between the velocity of the tip in UAD and the feed rate. The difference did not alter the contact ratio that could change the drilling force.

Figure 6.14 Influence of feed rate on thrust force in CD and UAD. Drilling speed \( N = 1800 \text{ rpm}, \) frequency \( f = 20 \text{ kHz}, \text{amplitude } a = 10 \text{ micrometers} \)
The effect of drilling speed on torque was also studied. In CD, the torque diminished significantly as the speed was changed from 600 rpm to 1800 rpm (see Figure 6.15). The effect was negligible for speeds above 1800 rpm. In UAD, the drop was linear up to the drilling speed of 2400 rpm and was not influenced by the speed above it. The average drop in the torque was about 30% in CD and UAD for the range of drilling speed used. The magnitude and behaviour of our results for drilling torque against drilling speed in CD was consistent with those reported by Hillery [67]. There it was shown that when using a 3.2 mm drill, there was an exponential falling-off of the cutting torque from 14.5 N.mm at 400 rev/min to an asymptote of 10 N.mm at 2000 rev/min.

![Figure 6.15 Influence of drilling speed on torque in CD and UAD. Frequency $f = 20$ kHz, amplitude $a = 10$ micrometers, $f_r = 40$ mm/min](image)

The influence of ultrasonic parameters on drilling thrust force and torque was studied. A similar behaviour for force decrease with increasing frequency and amplitude as obtained from the plane cutting experiments was observed in UAD. From Figure 6.16 and Figure 6.17, several trends can be deduced. Firstly, the drilling reaction torques and forces developed during UAD are lower than those generated during CD throughout the frequency range investigated. The thrust force decreased by 57% when the frequency was increased from 10 kHz to 30 kHz (Figure 6.16). The corresponding drop in torque was 28%. The force dropped by 46% when the ultrasonic amplitude was increased from 5 µm to 15 µm and was unaffected by a further increase (Figure 6.17). The torque diminished slightly (by 14%) when the amplitude was changed from 5 µm to 10 µm and remained unchanged under further increases in the amplitude. The drilling force has been shown to drop linearly with vibration amplitude in drilling ceramics using a diamond core drill.
Figure 6.16 Variation of thrust force and torque with frequency in UAD. Drilling speed $N = 1800$ rpm, $f_r = 40$ mm/min, amplitude $a = 10$ micrometers

Figure 6.17 Variation of thrust force and torque with amplitude in UAD. $N = 1800$ rpm, $f_r = 40$ mm/min, frequency $f = 20$ kHz

[163]. Here it is important to mention that the largest change in the thrust force and torque was for the maximum drilling speed of 3000 rpm. Different optimum values of vibration frequency and amplitude can exist for higher drilling speeds.

6.5 Mechanics of Bone Drilling

In drilling operations, small segmented chips are desirable. This is because as the chips get larger, they cannot move easily through the flutes, increasing the torque, specific cutting energy and possibility of bone thermal damage [48]. Increased cutting energy due to flutes clogging results in lower control by the surgeon of penetration through the bone. Also, excessive pressure on the drill may result in its slip, misalignment and risk of
breakage. Long bone chips can also cause geometrical inaccuracy and poor surface quality of a drilled hole that can lead to pin/screw loosening post-operatively. The study of drilling chip size and shape is important since it affects the temperature in the cutting zone, cutting energy and surface quality. Advanced understanding of the mechanics of bone splitting under the action of a vibrating drill is essential to improve drilling techniques.

An experimental analysis was performed to investigate the advantages of UAD as compared to conventional drilling (CD). This section studies mechanics of the bone drilling process with and without ultrasonic assistance, chip formation mechanisms and chip morphology which may affect the drilling thrust force and temperature in the cutting zone. A cylindrical column (with periosteum removed) from the middle diaphysis of a bovine femur as discussed in Section 6.3.1, was used in those tests.

### 6.5.1 Experimental equipment and conditions

A conventional lathe machine was used in the experiments with an ultrasonic transducer gripped in its chuck. Bone drilling parameters were kept constant in that series of experiments. In the ultrasonic regime, the drill vibrated with frequency of 20 kHz and an amplitude of 10 micrometers (Table 6.3). The cylindrical columns of cortical bone were held in a fixture, which was directly mounted on the feed bed. An experimental setup for drilling, high-speed filming and specimen fixation is shown in Figure 6.18. The tests were performed without cooling in the drill site. The drilling was conducted in the direction either parallel or perpendicular to the bone axis (i.e. osteonal direction) (see Figure 6.1). The drilling parameters in both CD and UAD are provided in Table 6.3.

![Figure 6.18](image-url) Experimental setup for bone drilling and high-speed filming
Table 6.3 Bone drilling parameters in CD and UAD

<table>
<thead>
<tr>
<th>Drill diameter D (mm)</th>
<th>Drilling speed N (rpm)</th>
<th>Feed rate $f_r$ (mm/rev)</th>
<th>Frequency $f$ (kHz)</th>
<th>Amplitude $a$ (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>1200</td>
<td>0.03</td>
<td>20</td>
<td>10</td>
</tr>
</tbody>
</table>

6.5.2 High-speed filming of bone drilling

Bone drilling is a fast process and it is difficult to visualise its mechanism without high-speed filming, which can provide direct observation of the drill-bone interaction zone. FASTCAM digital video recorder (Photron DVR, Photron Limited, Japan) was utilised for high-speed filming of the chip separation process (Figure 6.18). The camera was positioned on the tripod near the lathe and focused on the tip of the tool. The videos were recorded at 500 frames per second with resolution of $640 \times 480$ pixels for a duration of approximately 10 seconds. A macro zoom (18–108 mm) lens was used to zoom into the drilling location.

High-speed filming is capable of recording images of events occurring for short time periods and thus gives an opportunity to study the chip formation of a fast process such as drilling. One of the major limitations of high-speed filming is that recording at high frame rates requires the provision of very intense lighting and a proper focus on the area of interest. Two high-intensity lighting sets (quartz flood lighting) of 700 W were used to illuminate the filming area (Figure 6.18). The drilling speed and feed rate were kept constant in both conventional and ultrasonic drilling modes. Due to the light reflection from the tool and specimen, it was difficult to observe the cutting process on the screen. To eliminate this problem, the drill bit (tip and flutes) were blacked with a marker. That provided a clearer picture of the chip formation in the cutting zone.

6.5.3 Chip formation

At the first stage of experiments, a fresh cortical bone was drilled. The chip formation was observed from the moment the drill lips touched the bone surface till the end of the hole. In general, the chip morphology in drilling can be categorised with eight types: (1) needle, (2) powder, (3) fan, (4) short ribbon, (5) short spiral, (6) long ribbon, (7) long spiral, and (8) very long ribbon [164].

In CD of fresh bone in the direction parallel to osteons, the chips were found to follow a spiral path up along the flutes to the bone surface. That phenomenon continued until the
drill fully penetrated into the bone tissue. Continuous-mode cutting chips like those of a ductile material are generated for the small undeformed chip thickness using an endmill as reported by Mitsuishi [98]. The feed direction of the cutting tool in the experiment was perpendicular to the longitudinal direction of the bone. Clogging of the drill with generated chips started at the moment of full lip engagement and continued till the end of drilling the hole (Figure 6.19).

![Figure 6.19 Images obtained with high-speed video system showing chip formation in drilling fresh bone with both drilling modes. First row: CD, second row: UAD. Left column – initial engagement of drill lips; middle column – half-lip engagement; right column – full lip engagement](image)

The chips were seen rotating along the drill bit rubbing against the hole surface and blocking the flutes. The chips also impacted the hole wall surface which produced a bending moment in the chip leading to its fracture once the strain in the chip exceeded the critical value. Conventional drilling in the direction normal to osteons produced fan-shaped chips. These chips fractured prior to making a conical shape and drill flutes were clogged to a considerably smaller degree than in the previous case (i.e. parallel to osteons). Based on these observations, drilling in the direction parallel to osteons may require a higher penetration force as compared to that in the perpendicular direction as flute clogging results in the increased drilling force. Also the temperature rise in the cutting zone may be lower in the case of drilling in the direction normal to osteons as chips break quicker and do not stay in contact with the drill as long as in drilling in the direction parallel to bone axis.

UAD of fresh bone in both drilling directions resulted in needle-type segmented chips, and no significant differences in chip formation were noticed for various drilling directions. Flutes were observed to be clear from the chips during the drill advancement.
into the bone material. Also the chips were noticed to clear the hole with significant speed. Due to the small size and high speed of chip removal, a dust-like condition was observed in high-speed videos when ultrasonic drill touched the bone and continued until the full drill engagement (Figure 6.19).

The chips from all the experiments were collected in a tray and their morphology was further studied via microscopic examinations and is discussed in the next section.

**6.5.4 Chip morphology**

Optical microscopy of bone chips produced in the experiments described in the previous section was conducted. At the first instant, chips resulting from drilling of a dry bone were studied. These chips were mainly needle-shaped accompanied by traces of the fan-shaped ones. Chip shapes and sizes were almost similar for both drilling directions in the conventional drilling regime. UAD of dry bone produced only needle type chips for both drilling directions. CD of a fresh bone in the direction parallel to osteons produced predominantly short-spiral chips with traces of fan-shaped chips (Figure 6.20a). The length of spiral cone chips can be considered as a scale to evaluate the difficulty for chip evacuation in drilling [165]. The larger length of chips corresponds to the higher forces required for drill penetration due to flute blockage. CD in the direction perpendicular to the osteons.

![Image of chip shapes](image.png)

**Figure 6.20** Comparison of chip shapes generated by drilling fresh bone in both drilling regimes and directions: (a) CD parallel to osteons; (b) CD perpendicular to osteons; (c) UAD parallel to osteons; (d) UAD perpendicular to osteons
bone axis produced only fan-shaped chips (Figure 6.20b). A fan-shaped chip begins with a spiral form but does not curl sufficiently to follow the flute and hence fractures prior to a complete revolution. There is no clear demarcation between needle- and fan-shaped chips except that the latter curl slightly more and consequently have a larger surface area. UAD produced needle type chips accompanied by small chip fragments for both drilling directions (Figures 6.20c and 6.20d). In ultrasonic vibration-assisted drilling of the Inconel superalloy, the proportion of small size chips was observed to increase compared with the conventional technique [166]. The dust-like condition observed in the cutting zone in the high-speed filming experiment of UAD was due to the formation of such small “powder” chip fragments. The chip shapes produced in the experiments are summarised in Table 6.4.

Table 6.4 Chip shape produced by drilling for various directions and penetration modes

<table>
<thead>
<tr>
<th>Drilling type</th>
<th>Fresh bone</th>
<th>Dry bone</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Parallel to osteons</td>
<td>Perpendicular to osteons</td>
</tr>
<tr>
<td><strong>CD</strong></td>
<td>conical spiral</td>
<td>fan-shaped</td>
</tr>
<tr>
<td><strong>UAD</strong></td>
<td>Needle</td>
<td>Needle</td>
</tr>
</tbody>
</table>

The observations of the chip formation process in both drilling modes motivated the study of roughness of the drilled hole’s surface of the bone since length of a chip and the way it flows in drill flutes affect surface quality. The next section presents measurements of roughness of the drilled holes produced in both drilling techniques and their comparison.

6.6 Surface Analysis of Drilled Holes

Close tolerances and surface textures are critical for direct structural and functional connection between living bone and the surface of the implant. The character of the surface affects the bone-screw interface strength as well as the cellular response, which is essential for early and healthy bone growth. The quality of the bone-implant contact surfaces affects both contact stability and regeneration. The smoother the surface, the greater the extent of contact; whereas, the rougher the surface, the greater the stability [167]. Various techniques were employed to enhance bone apposition including bioactive coating and surface texturing of fixative components. Studies of surface roughening as a factor affecting the bone ingrowth have had encouraging results. There has been a continuing debate on the influence of implant surface topography on the success of operation. Internal fixation screws require a stable bone-implant interface for transmission of forces [168]. On
the other hand, strong integration between the bone and screw is a disadvantage when considering removal of screws (especially in paediatric patients with fast growing bones), and the surface microstructure is the main determining factor here. Bone integration is minimised by using surfaces with minimal microstructure reducing the forces required to remove screws. Numerous studies investigated the effects of surface texture or microtopography on interfacial strength and the cellular response, both in-vivo and in-vitro (see e.g. [169][170][171]).

Previous research was mainly concerned with the measurement of the implant’s surface topography in relation to anchoring to the bone. To authors’ knowledge, there is no study reported so far, describing the bone hole surface roughness produced with either CD or UAD. Evaluation of the contact between the bone and fixative components (i.e. screws or pins) that takes into account the bone surface roughness would allow more precise analysis tools to be built. It is also envisaged that an experimental study will be conducted in the future to investigate the influence of the bone surface roughness on the fixation strength. This study is concerned with measurements of surface roughness of holes drilled in a cortical bone using two studied drilling techniques. The hole surface roughness produced with conventional drilling (CD) and ultrasonically-assisted drilling (UAD) was measured with, and compared for, contact and non-contact methods. The present study is aimed at measurement of the average roughness of the drilled hole’s surface, which is by far the most extensively used surface parameter.

6.6.1 Non-contact roughness measurements

Surface roughness is the measure of fine surface irregularities in the surface texture. These are the result of the manufacturing processes employed to create the surface. Surface roughness parameter ($R_a$) is rated as the arithmetic average deviation of the surface valleys and peaks expressed in microinches or micrometers. Two different methods were used to characterise roughness of surfaces of drilled bones. The Talysurf CLI 2000 system was used in our non-contact analysis; this is a scanning topography measurement instrument (experimental arrangement is shown in Figure 6.21). It means that the gauge measures the altitude of only one point at a time, and the study sample is moved on the cross-slides in order to scan the complete measurement area. The confocal point gauge 300 based on
chromatic length aberration (CLA) principle was used in our high-resolution measurements. In this technique a white light is directed by a beam splitter through a spectral aberration lens onto the surface. The lens splits the light into different wavelengths and at any point on the surface only a certain wavelength is focused. Light is reflected from the surface to a pin hole, which permits only the wavelength in focus to pass through. A spectrometer deflects the light onto a CCD sensor to interpolate the spatial position of the data point. The instrument can scan areas up to 200 mm × 200 mm with maximum measurement speed 30 mm/sec. The specifics of this test in measuring bone surface roughness will be given in the section with results.

6.6.2 Contact roughness measurements

The Talysurf 4 system shown in Figure 6.22 was employed in contact measurements of the roughness of drilled holes. It is a surface texture measuring instrument with a
stylus traversed across the surface; its vertical movement is converted into an electrical signal. The $R_a$ value is derived from the filtered signal and is displayed on either a pointer or digital type meter. The pick-up used in Talysurf 4 is based on a variable position-sensitive inductance, with a signal proportional to the displacement, even when the stylus is stationary. Styluses of various sizes can be fit on the system to handle a variety of sample sizes and shapes.

### 6.6.3 Experimental procedure

Ten holes were drilled with each drilling technique – CD and UAD. The drilling speed and drill diameter used in the experiments (see Table 6.5) were chosen according to the data widely reported in literature for bone drilling [67][78]. The feed rate and drilling speed were kept constant throughout the experiments. Drilling was performed in a direction perpendicular to the cortical wall and the bone axis. After drilling, the parts of the bones with holes were cut with a hacksaw along the axis of the bone (line AB in Figure 6.23) to provide an access to the formed surfaces for measurements. Test samples each containing two holes drilled with CD and UAD, respectively, were then cut off from the original sample for examination. The transverse edges of the samples were polished to remove burrs thus leaving 6 mm hole depth (in the axial direction of drilling shown in Figure 6.23) for examination.

The $R_a$ values were measured using two techniques discussed above. In a non-contact method the amplitude for the points of 2D profile of the surface was captured with the CLA transducer (resolution of 0.02 nm). The spatial resolution for the stage movement was 0.5 µm beneath the CLA gauge. The specimens were taken for surface examinations immediately after drilling to avoid the effect of drying. The surface roughness measurements of the hole surface were taken along the axis of the drilled hole (line EF in Figure 6.23). The surface roughness was calculated using the relation

$$ R_a = \frac{1}{L} \int_0^L |f(x)| dx \tag{6.4} $$

where $L$ and $x$ are the evaluation length and the distance along measurement, respectively. Drilling parameters used in these experiments are provided in Table 6.5.
Table 6.5 Drilling parameters used in experiments

<table>
<thead>
<tr>
<th>Parameters</th>
<th>CD</th>
<th>UAD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Drilling speed (rpm)</td>
<td>2000</td>
<td>2000</td>
</tr>
<tr>
<td>Drill diameter (mm)</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>Feed rate (mm/min)</td>
<td>40</td>
<td>40</td>
</tr>
<tr>
<td>Vibration amplitude (µm)</td>
<td>N/A</td>
<td>10</td>
</tr>
<tr>
<td>Vibration frequency (kHz)</td>
<td>N/A</td>
<td>20</td>
</tr>
</tbody>
</table>

Figure 6.23 Stages in sample preparation and direction for roughness measurements

6.6.4 Results and discussions

**Non-contact method**

Bone samples were mounted on the cross-slides of the Talysurf CLI 2000 instrument. The scanning speed was kept at 0.5 mm/s. The instrument scanned an area of 6 mm × 4 mm along the entire hole depth. The distance between the sample and scanner was found to be approximately 6 cm to achieve better surface intensity for measurements. The scanned images were analysed using TalyMap software. An oblique plot of the scanned area is shown in Figure 6.24. The software has the capability to remove the cylindrical effect from the scanned area and convert it into a flat one for measurement purposes. A typical surface profile obtained with Talysurf CLI 2000 is shown in Figure 6.25 while Figure 6.26 provides a comparison of $R_a$ values for CD and UAD. The average levels of $R_a$ for 10 holes drilled with CD and UAD were 1.52 µm and 1.35 µm, respectively. It means that the hole surface obtained with CD was rougher than that drilled with UAD.
Figure 6.24 Oblique views showing roughness details on the surface obtained using Talysurf CLI 2000 with cylindrical effect (a) and after cylindrical effect was removed (b). Scanned area 6 mm × 4 mm

Figure 6.25 Roughness pattern obtained with Talysurf CLI 2000 along hole depth

Figure 6.26 Comparison of roughness parameter $R_a$ of hole surface using Talysurf CLI 2000. Each bar graph for each sample represents the mean of five scans
Contact method

The non-contact method applied required a sufficiently reflective surface for accurate measurements [172]. The measured bone surface was light in colour, and it was very hard to focus on it for examination. Hence, to validate the results obtained with the non-contact method described above, the contact method was also used. In that method, a physical contact between the surface and a measuring probe (stylus) was achieved. The sample was carefully mounted on the stage of Talysurf 4. The stylus was moved over the surface with a speed of 10 mm/min. A typical surface profile obtained with Talysurf 4 is shown in Figure 6.27; a comparison of measured $R_a$ values for CD and UAD is given in Figure 6.28.

![Graph showing roughness pattern obtained with Talysurf 4 along hole depth](image)

**Figure 6.27** Roughness pattern obtained with Talysurf 4 along hole depth

The average $R_a$ values for 10 bone specimens drilled with CD and UAD was 1.37 µm and 1.28 µm, respectively. Those values are in a good agreement with the magnitudes provided by the non-contact method. The effect of drilling technique on quality of the hole

![Graph comparing roughness parameter $R_a$ values of hole surface using Talysurf 4](image)

**Figure 6.28** Comparison of roughness parameter $R_a$ values of hole surface using Talysurf 4. Each bar graph of each sample represents the mean of five measurements.
surface may be explained by observing the process of bone drilling. To study the interaction between the bone and the drilling tool, chip formation mechanisms were studied using high-speed filming of the drilling process. The main stages of these processes are presented in Figure 6.19 (Section 6.5.3).

It was observed that CD produced spiral cone chips while UAD produced needle-shaped chips. In CD the chips were seen rotating along the drill bit rubbing against the hole surface and blocking the flutes. UAD was observed to produce broken chips and no rubbing of the chips against the hole surface was seen. Such spiral chips were reported to cause higher specific cutting energy and rough hole surfaces in drilling cast aluminium alloys [173]. Based on these observations, it was concluded that the surface roughness produced by UAD should be lower than that in CD, which was consistent with experimental results for surface roughness.

6.7 Concluding Remarks

An experimental programme was undertaken to measure the tool penetration forces with and without ultrasonic assistance/vibrations for two types of cutting i.e. plane cutting and drilling. The chip formation mechanism, which affects cutting forces and temperature in the cutting zone was studied for conventional drilling (CD) and ultrasonically-assisted drilling (UAD). The effect of a vibrating tool on the hole surface quality which influence bone-implant interfacial strength was also studied.

Experimental results revealed that the penetration force in both types of cutting was dropped significantly when ultrasonic vibrations were imposed on the tool. The drop was about 50% when the DOC was smaller (0.1 mm) in plane cutting. Similarly, the drilling force was nearly halved in the presence of vibrations for the range of drilling speeds used. The cutting force decreased linearly with ultrasonic frequency used while there was a limit for the amplitude (15 micrometers), above which the decrease in the force was insignificant for both types of cutting. The drilling torque also diminished by a considerable amount in the presence of drill vibrations reducing the chance of drill breakage. The decrease in the force will allow the surgeons to penetrate with the tool with less power and to have a better control over it. This will also minimise the risk of damage caused by the tool load.

The mechanics of chip formation and chip morphology were examined in this study for both conventional drilling (CD) and ultrasonically-assisted drilling (UAD) applied in two drilling directions: perpendicular to the osteonal direction (or bone-axis direction) and parallel to it. CD parallel and perpendicular to the osteonal direction produced spiral cone
and fan-shaped chips, respectively, which showed that chip size and morphology are direction-dependent. The chip formation mechanisms observed in the high-speed filming experiment for UAD of fresh bone were similar to that for CD of dry bone. UAD of fresh bone in both drilling directions produced needle-shaped chips similar to those in CD of dry bone.

An advantage of UAD compared to CD was an improved chip removal from the drilling site. High-speed filming and the resulted chip morphology showed UAD to be a better choice for drilling fresh bone than CD as it produced segmented chips and facilitated fast chip evacuation from the drilling zone, thus potentially reducing undesirable heat build-up during drilling. Spiral chips and their slow evacuation in CD would result in a higher bone temperature potentially leading to bone necrosis. Chip formation and morphology observed in these experiments enhance our understanding of the mechanism of bone drilling. These results will be used in future computational modelling of the bone drilling process.

The levels of surface roughness of holes drilled in the bovine cortical bone using two drilling techniques were measured. The surface texture indicated that the hole surface produced by ultrasonically-assisted drilling was somewhat smoother than that obtained with the conventional one. The difference in roughness was about 6.5% as measured with the contact method and 11% with the non-contact one. The intermittent contact between the drill and bone in UAD reduces the mean coefficient of friction which was responsible for the smoother surface. The values of $R_a$ obtained with both measurement methods were in good agreement. Hence, non-contact methods may be confidently applied to measure the surface quality of bone tissues. The non-contact method was relatively quick and handled larger areas. Lower surface roughness produced in UAD may be attributed to an improved chip removal from the drilling site. A rough surface may be achieved using CD, which will provide better anchorage of bone tissue to the implants and fixing screws. For a reduced drill penetration force and improved chip removal to avoid flutes blocking and risk of drill breakage, UAD would be a better choice.

Further research is needed to evaluate the integrity of the drilled hole surface, in order to ensure the interfacial strength is not affected. The selection of the optimum drilling technique for clinical practice is a complex matter and depends strongly on the objectives of the operation. Ultrasonic drilling techniques currently have limited applications in orthopaedic surgery, and this study is hoped to be a step towards getting better understanding and subsequent optimisation of such techniques in order to facilitate their wider use. However, the additional temperature load imposed by the high-energy
vibrations into the cutting zone still requires future investigation. The analysis of mechanics of cutting and its effect on bone tissue would facilitate the design and fabrication of advanced surgical instruments.

The chip formation mechanism and cutting force strongly influence the temperature at the cutting zone in all cutting processes particularly in drilling. As drilling is the most frequently applied surgical method in orthopaedics, a detailed analysis of the temperature rise in bones is necessary to investigate the cutting conditions inducing thermal necrosis. These measurements will be used to validate our results obtained with the finite element model of bone cutting. In the next chapter, bone drilling temperatures are measured and analysed using thermocouples.
Chapter 7. Temperature Measurements in Drilling Cortical Bone

7.1 Introduction

Heat generation is a well known problem in drilling of bone due to its sensitivity to rises in temperature beyond a certain threshold level. In drilling, chips flow in flutes of the drill bit, obstructing the conduction of heat away from the cutting zone due to lower conductivity of the bone. Temperature measurements are of fundamental importance in orthopaedic bone drilling, since exposure to elevated temperatures may induce irreversible thermal damage. Temperature has a harmful effect on the bone tissue because it is directly linked to the thermal damage/necrosis of bone associated with irreversible changes in its structure and physical properties. If necrosis sets in, it can break down the bone around the implantation site, causing loosening of fixtures (screws, pins) and can dramatically weaken the whole structure.

Mechanical rotary drills are the main type used in orthopaedic surgeries. The analysis of temperatures during a high-speed cutting process like drilling is necessary for investigating the level of temperature rise in the bone tissue. Such findings may be helpful in designing the surgical equipments in order to minimize the possibility of bone thermal damage. Previous studies, while measuring drilling temperatures, used drilling perpendicular to the bone princible direction, and the penetration depth was limited to the wall thickness of cortical bone (5–9 mm) [67][78][79]. It is not always the case and the penetration depth can be higher, e.g. in Anterior Cruciate Ligament (ACL) reconstruction surgery [174].

Previous studies were mainly focused on measurements of drilling temperatures using either bone-imbedded or tool-imbedded thermocouple systems. There is no study that has compared temperature measurements with both measurement systems. Further the drilling temperature were measured in drilling through a single cortical wall thickness. This chapter presents at first an analysis of temperatures measured in bone drilling with a penetration depth equal to a single wall thickness followed by a study to acquire a saturation temperature for a drilling depth higher than the thickness of a single cortical wall. The aim was to find critical drilling parameters when subjecting bone to higher temperatures. A detailed experimental programme was undertaken to analyse the bone temperatures using tool-imbedded and bone-imbedded thermocouple systems and drilling both a single cortical wall thickness (7–10 mm) and penetrations up to 40 mm. Temperatures were measured in conventional drilling (CD) only due to interaction of thermocouples with ultrasonic drill in both measurement systems.
7.2 Heat Generation in Bone Drilling

Heat generation in cutting materials is mainly due to the rate of deformation as well the friction between the tool and material being cut. In the cutting process, the maximum heat is generated at the plane of the contact between the tool and the work piece and is given by Davidson [75]

\[
\frac{\partial Q}{\partial t} = F_s v_s
\]  

(7.1)

where \( Q \) is the heat generated, \( t \) is time, \( F_s \) is the shearing force in the shear plane, \( v_s \) is the shear velocity in the shear plane. The amount of heat penetration at either side of the contact is strongly influenced by the material property called the coefficient of heat penetration or thermal effusivity. If two materials of different effusivities are in thermal contact, the amount of thermal flux into either of the materials will depend on the value of the effusivity of each material [175]. The next section describes the measurement of the temperatures during bone drilling using thermocouples.

7.3 Tool-imbedded Thermocouple Measurements

Limited studies have reported temperature measurements in the cutting process using a thermocouple imbedded in the tool. To our knowledge, there is only one study where a tool-imbedded thermocouple with slip rings arrangement was used to measure temperatures in bone drilling [67]. The average temperature, at which the necrosis most likely appeared, was reported at a threshold around 50-70°C [70][72][73]. In this study, a threshold level for thermal necrosis was chosen as 70°C as reported by Moritz [70]. Currently, there is no study available reporting the temperature of the bone exposed for surgical incision and therefore, in this study, the initial temperature of the bone was considered as 25°C (the initial temperature of bone measured before the experiments). The value on the ordinate, \( \Delta T \) of the subsequent plots represents the rise in temperature above 25°C. Therefore the necrosis level was considered when the rise in bone temperature (\( \Delta T \)) attained the value of 45°C. It is worth mentioning that this level was based on the initial temperature of 25°C and may be different for other experimental (initial) conditions.

7.3.1 Specimen preparation

Drilling tests were performed on specimens excised from bovine femur used in experiments of plane cutting and drilling in Chapter 6. The average thickness of cortical wall was 10 mm. Two types of specimens were prepared: (1) with a single cortical wall
thickness and (2) with multiple cortical wall thickness to achieve larger drilling depths. In the later case, specimens with single wall thickness were cut with a rotary hacksaw and flattened along the longitudinal axis using a milling machine to then be placed on each other. A thin layer of glue was applied between the smooth surfaces to provide a strong contact. The anatomical location and dimensions of the specimen are shown in Figure 7.1. The final dimensions of the specimen were 50 mm × 40 mm × 40 mm.

Figure 7.1 (a) Fresh bovine bone; (b) its sample cut from mid-diaphysis; (c) arrangement for larger drilling depths

7.3.2 Experimental equipment

A conventional lathe machine was used in these experiments with a bone specimen gripped in its chuck and a drill fixed in the tool post. A standard twist drill (5 mm diameter) made of tungsten carbide and carrying a hole (originally manufactured for coolant purpose in metals drilling) along the drill flutes was used in the tests due to its high thermal conductivity. The purpose of using a drill with high thermal conductivity was to measure the temperature close to the cutting edge. The coolant hole diameter was 1 mm and was at a distance of 1.5 mm from the drilling lip/cutting edge. Standard K type insulated thermocouples with a wire diameter of 127 μm were used for temperature measurements. The K-type thermocouple can take measurement values up to 500°C with a response time of 10 μs. At the beginning of each set of test, a calibration procedure of the thermocouples was performed against a laboratory thermometer within ±0.3°C. For temperature acquisition, data logger FMSDL48 (Glasgow, UK) was used. Before the experiments an instrument calibration of the entire system was performed against the mercury thermometer. A system error of approximately ±1°C was found for temperature
measurement. To ensure good contact of thermocouple with drill, their beads were coated with thermo-conductive paste (Omega, OT-201-2). The drill-bit thermocouple’s location and experimental setup for drilling is shown in Figure 7.2 and Figure 7.3.

Figure 7.2 Schematic of drill design and thermocouple location

Figure 7.3 Experimental arrangement for drilling bone with drill-imbedded thermocouple system

7.3.3 Experimental method

Bone drilling parameters were changed in the experiment to investigate their effect on the temperature rise in bone: their values are given in Table 7.1. Twenty holes were drilled in each specimen. All the experiments were performed without irrigation (cooling) at the drill site. Drilling was conducted in the direction perpendicular to the bone axis (i.e. normal to osteonal direction). The initial temperature of bone and drill was measured and was found to be 25°C. The drill was held stationary due to the presence of thermocouple
wire and the bone specimen was rotated in the chuck. The feed was provided by the drill. The diameter of the drill used in orthopaedic bone drilling is generally 2–4 mm with rotational speed ranging from 1000 rpm to 3000 rpm [67][78][79][81]. Arbitrary values of feed rate were chosen in the experiments as the penetration speed of the hand-held drill in bone varies from surgeon to surgeon in clinical practice. Each data point on the subsequent plot represents the average value of five tests.

**Table 7.1** Parameters used in temperature measurements in drilling

<table>
<thead>
<tr>
<th>Drill diameter (mm)</th>
<th>Drilling speed (rpm)</th>
<th>Feed rate (mm/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>4, 5</td>
<td>600 – 3000</td>
<td>10 – 50</td>
</tr>
</tbody>
</table>

**7.3.4 Drilling speed vs. bone temperature**

To account for the change in bone temperature with drilling speed in specimens with a single cortical wall thickness, they were rotated with speed ranging from 600 rpm to 3000 rpm for a constant feed rate of 50 mm/min. Temperature values at each drilling speed were recorded when the drill tip just exited the cortical wall (approx.10 mm). The effect of drill speed on temperature rise in the bone is shown in Figure 7.4. Each data point in Figure 7.4 represents the maximum temperature obtained just before the drill tip exited the cortical thickness. The temperature was observed to drop rapidly as soon as the drill exited the cortical wall. It can be seen that the dependence of temperature on drilling speed is rather significant, since a temperature increase of 38°C was recorded when changing the drilling speed from 600 rpm to 3000 rpm. The rise in temperature was due to the fact that as the drilling speed increased, the rate of deformation in the material and the friction between the drill and bone also increased, and so does the speed of heat flux generated. The increase in temperature at higher drilling speed caused the thrust force to drop and was explained in Section 6.4.3. The increase in temperature was sharp up to a drilling speed of 1800 rpm, and the rate of increase dropped when the drilling speed was further increased. An increase in thermal conductivity of bone with increasing temperatures as reported by Moses [51] and Wipf [52] may be the reason for producing heat at lower rate at higher drilling speeds. A similar behaviour of temperature rise has been reported in drilling human and bovine cortical bone [67] and AISI 1045 steel [176] with drilling speed exceeding 1200 rpm. An identical effect of cutting speed on the tool-shim interface temperature was studied in an experimental and numerical investigation of cutting titanium
alloys [177]. For a single cortical wall thickness/hole depth tested in drilling, the temperature was not seen to tend towards saturation point (Figure 7.4).

### 7.3.5 Influence of feed rate on temperature

The feed rate is one of the key parameters influencing drilling force and temperature and is directly linked to the operation/osteotomy time. The response of a thermocouple as a function of drilling time at three drilling speeds and two feed rates is highlighted in Figure 7.5. Due to higher velocities the chip developed more quickly resulting in higher temperatures at the cutting zone. By comparing Figure 7.5a and Figure 7.5b, it is evident that the temperature developed more quickly in the case of the higher feed rate for the same drilling speed. The temperature was observed to rise nearly linearly from the time when the drill touches the top surface of the bone and then diverted from it with a lower slope. This may be explained in terms of mechanics of drilling by differences between the two stages, i.e. drilling up to a full lip engagement between the drill and bone and when it penetrated further. The temperature was observed to drop quickly as the drill tip exited the cortical thickness (not shown in Figure 7.5). Similar to the previous results, the temperature rise was not seen to saturate when drilling through a single cortical thickness.
7.3.6 Acquiring saturation temperature

In anterior cruciate ligament (ACL) reconstruction surgery, the drill has to penetrate into the femur up to 30 mm depth with drill diameter 9–10 mm [174]. Also when drilling parallel to the longitudinal axis of bone, the drill may go longer than the average single cortical wall thickness. To our knowledge, no research has been carried out previously to investigate the magnitude of the drilling temperature rise for a penetration depth larger than a single cortical wall thickness. It is obvious from Figure 7.5 that if there was more penetration depth available for the drill, the temperature would have been increased further. The objective of the present study was to investigate the depth, at which the drilling temperature saturates. Experiments were conducted to measure the level of maximum temperature (saturation temperature) by drilling thick specimens through the
entire thickness of 40 mm. Figure 7.6 shows the variation of temperature with time for a constant drilling speed of 1800 rpm and two values of feed rates. With a higher feed rate, saturation temperature was achieved more quickly (35 seconds at 50 mm/min) and was relatively higher than that obtained with a lower feed rate (90 seconds at 20 mm/min). The corresponding measured saturation temperature increase was 120°C and 90°C for the two feed rates used.

![Figure 7.6](image)

**Figure 7.6** Temperature evolution in drilling thick specimens with two different feed rates, $N=1800$ rpm

For a feed rate fixed at a level of 50 mm/min, the temperature variation at various drilling/penetration depths using the range of drilling speeds provided in Table 7.1, is shown in Figure 7.7. The higher the drilling speed, the larger the increase in bone temperature. It is obvious that as the drill-bit travels further into the bone there is an increase in temperature due to the inability of the bone to conduct the heat generated away from the drilling site. Temperature was found to saturate between the hole depth of 30 mm and 40 mm for all drilling speeds used.
The effect of feed rate on a rise in temperature at various drilling/penetration depths is shown in Figure 7.8. A linear relation was obtained between the temperature rise and feed rate at four penetration/drilling depths. The increase in temperature may be caused by the increase in the amount of passing materials per unit time. A similar relationship between the drilling temperatures and feed rate for AISI 1045 steel was found by Ueda [176]. An identical behaviour of the temperature with feed rate at the tool-shim interface was reported in turning of titanium alloys [177]. For drilling through a single wall thickness, the rise in temperature was changed from 68°C to 93°C (increased by 37%) when the feed
rate was varied from 10 mm/min to 50 mm/min. An increase of the same magnitude was measured for the same range of feed rates at the drilling/penetration depth where temperature was stabilised (between 30 mm and 40 mm). Our results contradict those measured in drilling bone [178] where a decrease in temperature with an increasing feed rate per tooth (ranging from 0.02 mm to 0.1 mm) was measured. In that study, the decrease in temperatures was attributed to the short time exposure of bone to drill with increased feed rate.

7.4 Bone-imbedded Thermocouple Measurements

In this section the temperature measured by inserting thermocouples in bone is analysed. The objective was to compare temperature rise in bone measured by two systems – bone-imbedded thermocouple system and the tool-imbedded system. Similar to our previous experiments, specimens with single and multiple wall thickness were drilled. To imbed thermocouples, holes of 1 mm diameter were drilled in the bone. For temperature measurements in drilling through single wall thickness (10 mm), a thermocouple was placed about 7 mm from the top surface with its bead 1 mm from the drill cutting edge (see Figure 7.9(b)). For temperature measurements at higher drilling depths, thermocouples were placed 7 mm apart from each other along the drilling path as shown in Figure 7.9(b). With this arrangement, five thermocouples were placed along the drilling track (40 mm) to measure temperature. Thermocouples used in the previous experiments were imbedded for temperature measurements. All thermocouples were placed at a distance of 1 mm from the drilling tract, in order to avoid crushing by the propagating drill. Drilling tests were performed using a vertical drilling machine. An experimental setup for bone drilling with bone-embedded thermocouple is shown in Figure 7.9(a).

Figure 7.9 Experimental arrangement for drilling bone with bone-imbedded thermocouple system: (a) bone with thermocouples; (b) schematic of thermocouple’s locations along drilling depth
No significant temperature rise was recorded until the cutting edge approached the location where the thermocouple was placed. The temperature rose very quickly up to a maximum value and decreased as the cutting edge passed the thermocouple bead as shown in Figure 7.10.

![Figure 7.10](image)

**Figure 7.10** Typical response of thermocouple placed in single wall thickness at depth of 7 mm. $N = 1800$ rpm; $f_r = 50$ mm/min

Similar to the tool-imbedded thermocouple system, temperature measurement was strongly influenced by the drilling speed as shown in Figure 7.11. The bone temperature

![Figure 7.11](image)

**Figure 7.11** (a) Temperature rise vs. drilling speed using bone-imbedded thermocouple system in drilling through single wall thickness, $f_r = 50$ mm/min (b) experimental arrangement for drilling through single wall thickness with bone-imbedded thermocouple system.
was increased with increase in the drilling speed. The temperature rise was increased from a mean value of 6°C to 51°C when the drilling speed was changed from 600 rpm to 1800 rpm and the relationship was linear. A further increase of 16°C was measured between 1800 rpm to 3000 rpm. The drop in increasing rate at higher drilling speed was similar to that obtained with the tool-imbedded thermocouple system. The results of temperature measurements in these experiments contradicted those reported by Abouzgia [80], where a non linear decrease of bone temperature was reported with increasing drilling speed. The obvious reasons responsible for those differences were the use of 2.5 mm diameter drill and significantly higher speeds (from 20000 rpm to 100000 rpm) in that study.

In the next set of experiments, the drill was penetrated through the entire depth of thick specimens with three drilling speeds, and temperatures were recorded at each thermocouple location. It was not worthy to establish the relationship between a single thermocouple reading and time due to quick temperature rise up to a maximum value followed by its sharp drop as the drill approached and passed the thermocouple location. The data points in Figure 7.12, represents thermocouple measurements at locations placed along the drilling path (see Figure 7.9b). The temperature was found to saturate for the drilling time above 35 seconds for three drilling speeds, which correspond to the drilling depth between 30 mm and 40 mm for a feed rate of 50 mm/min. This behaviour was consistent with those obtained using tool-imbedded thermocouple system.

![Figure 7.12](image)

**Figure 7.12** Temperature measurements obtained with five thermocouples inserted in bone sample at various depths from the top surface. $f_r = 50$ mm/min
7.5 Conclusions

The bone drilling experiments allowed us to quantify temperature levels in bones using tool-imbedded and bone-imbedded thermocouple systems and to compare them. A number of parameters such as drilling speed, feed rate and drilling depth were identified to affect the drilling temperatures. Also the level of measured temperatures in bone was observed to be influenced by the method applied for such measurements. The maximum temperature obtained from tool-imbedded thermocouple system bone was higher than previously reported due to the measurement limitations of previous studies. The temperature rise as measured with tool-imbedded thermocouples was above the thermal threshold value ($\Delta T = 45^\circ C$) for all the drilling speeds and feed rates that could produce damage to the bone as described in the previous studies. Only drilling through a depth lower than the average/single wall thickness, without irrigation, would avoid necrosis i.e. when the drilling time was below 4 seconds and 12 seconds, respectively for the feed rate of 50 mm/min and 20 mm/min using maximum drill speed of 3000 rpm. These times correspond to the drilling depth of approximately 4 mm and 3.5 mm using feed rates of 20 mm/min and 50 mm/min and maximum drill speed of 3000 rpm. The systems measured the saturation temperature at a depth above 30 mm for all drilling speeds and feed rates. The method was useful in measuring the evolution of the temperature rise which may be helpful in the analysis of bone necrosis using analytical methods involving temperature-time relationships.

Bone-imbedded thermocouple systems measure lower temperatures in bone for the similar drilling conditions used in tool-imbedded systems. This can be explained by the presence of a thermal barrier of approx. 1 mm between the drill and the bead of the thermocouple. The difference may also be due to the porosity in the bone microstructure (e.g. canaliculi and haversian canals) reducing the area for heat transfer. The technique however, identified parameters influencing drilling temperatures in the same fashion as did by the tool-imbedded thermocouple system. According to this technique, necrosis will not occur even at larger drilling depths if the drilling speed was kept equal or below 600 rpm. Both measurement techniques identified the same drilling depth where temperature was saturated. This study suggests that lower levels of the drilling speed and feed rate for bone surgeries could decrease the thermal damage to bone and prevent necrosis. To minimise thermal invasion, appropriate cooling conditions should be used. Additional experiments are needed to investigate the drop in the drilling temperature for cooling conditions at the drilling site. As drilling speed was identified to be the most influencing parameters in
producing higher temperature in bone, therefore, the use of high drilling speed in orthopaedics should be the subject of future research.

In the next chapter a finite element bone cutting model is developed for the purpose of calculating the tool penetration force and temperatures at the cutting zone. These experimental results for temperature measurements will be used to validate our FE bone cutting model with respective discussion in Chapter 10.
Chapter 8. Development of Finite Element Model

8.1 Introduction

Finite element modelling is by far the most useful tool to simulate the process zone and tool-work interaction in material cutting. Despite the modelling capabilities of finite element methods, it has only a few applications in modelling the process of cutting biomaterials. To improve bone cutting methods, optimization of cutting parameters is necessary. Presently, this requires expensive experimental equipment and additional safety measures to protect from biohazards. Finite element (FE) modelling of bone cutting could be an important extension to, and possible substitute for, experimental studies, reducing relating high costs well as potential health risks associated with biological materials.

A FE analysis of bone cutting has not been reported in literature so far; this may be caused by unavailability of correct material properties for such modelling and difficulties inherent to numerical modelling of material separation. The studies performed are limited to modelling of bone (cortical and trabecular) orthotropic/anisotropic material properties, bone damage, bone remodelling and stress and strain estimations under various loading conditions. For years, researchers have employed mechanical testing and finite element (FE) models to study the fracture process in bone under various loading conditions. The numerical studies have been limited to FE cutting models for materials similar to bone such as E-glass filled epoxy resin, polyurethane foam and wood [8][137]. Despite the reported application of UAC in bone cutting [9][87], the process was not modelled using numerical techniques. The influence of UAC on the material behaviour in bone cutting using FE simulations has not been studied yet.

This chapter presents the development of two-dimensional thermomechanically coupled FE models of both conventional cutting (CC) and ultrasonically-assisted cutting (UAC). The models incorporate a description of the advanced material behaviour, taking into consideration its non-linear hardening and strain-rate sensitivity effects of the material during cutting. They also take into account various heat-transfer, friction and cooling conditions at the chip-tool interface. For UAC, the vibration motion is considered in the suggested model to be in the cutting direction, with the cutting tool vibrating over the surface of the work piece. Direction-dependent material properties of the bone are idealised using the concept of isotropy and equivalent homogeneous material (EHM). The following section discusses a detailed scheme for the development of a FE bone cutting model.
8.2 Selection of Process Parameters

Mechanical drills are the most common tools used in bone surgery. The drilling process has a very complex kinematics but it may be approximated as a 2D orthogonal cutting process at the outer corner of the cutting lip [132][139]. Although the cutting action in drilling and orthogonal cutting is different, the basic mechanism of chip removal is the same and thus the theory for orthogonal cutting is still applicable. The schematic of this approximation is shown in Figure 8.1.

![Figure 8.1 Cutting lips of twist drill (a) with application of orthogonal cutting model where the orthogonal cutting model (b) (after [132])](image)

The drilling diameter and speed used in previous chapters for the measurements of the drilling force and temperatures are also considered here. The range of the tool’s cutting speeds ($V_c$) for simulations was calculated as a conversion of the rotational speed at the outer corner of a 4 mm diameter drill into a linear one for the drill speed up to 3000 rpm:

$$V_c = \frac{\pi DN}{60},$$

(8.1)

where $V_c$ is the linear cutting speed, $N$ is the rotational speed in rpm and $D$ is the diameter of the drill. As the exact frictional condition between the tool and bone was unknown, the value of friction coefficient (0.35) was taken from [Wyeth-2007][179] due to the similarity in the chip formation behaviour for bone and wood. Also the range of cutting force in cutting wood reported by [Sinn-2005], for the same chip thickness used in plane cutting experiments (see section 6.3.4), was close to our experimental results. The study of the effect of friction on bone cutting is given below, Section 9.3.1.
The value of the cutting edge angle and rake angle of an orthopaedic drill were measured and found to be 65° and 20°, respectively, and were used in modelling the cutting tool. The nose radius of the sharp edge of the tool used in plane cutting experiments was defined from a microscopic image as approximately 10 micrometers. The maximum value of DOC (0.3 mm) was chosen in plane cutting experiments as above that value the generated chips were no longer continuous. It is worth mentioning that the DOC produced by the five clinicians was between 0.1 mm and 0.3 mm in studies into the level of force produced for bone cut with a chisel and conventional bur [180]. In numerical simulations, parameters such as the DOC, nose radius (ψ) and cutting velocity were changed to observe their effect on cutting forces.

The normal core body temperature of a healthy, adult human being is stated to be 37°C. Hence, in all simulations the initial temperature of the work piece was assigned 37°C to present in-vivo bone temperature. The thermal threshold level for inducing necrosis was considered 70°C as used in bone drilling temperature tests. The cutting parameters together with their magnitudes used in simulations are provided in Table 8.1

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Magnitudes used in FEA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cutting speed, ( V_c ) (mm/s)</td>
<td>100; 200; 300; 400; 500; 600</td>
</tr>
<tr>
<td>Friction coefficient, ( \mu )</td>
<td>0.35</td>
</tr>
<tr>
<td>Tool nose radius, ( \psi ) (µm)</td>
<td>0; 5; 10; 15</td>
</tr>
<tr>
<td>Tool rake angle, ( \alpha ) (deg)</td>
<td>20</td>
</tr>
<tr>
<td>DOC (mm)</td>
<td>0.1; 0.15; 0.2; 0.25; 0.3</td>
</tr>
<tr>
<td>Amplitude (µm)</td>
<td>10; 15; 20; 25; 30</td>
</tr>
<tr>
<td>Frequency (kHz)</td>
<td>10; 15; 20; 25; 30</td>
</tr>
<tr>
<td>Bone initial temperature (°C)</td>
<td>37</td>
</tr>
<tr>
<td>Convection heat transfer, ( h ) (W/m²K)</td>
<td>0; 8000</td>
</tr>
</tbody>
</table>

8.3 Bone as an Equivalent Homogeneous Material

Cortical bone, with its osteons aligned predominantly along its longitudinal axis, is a heterogeneous material and can be approximated as a fibre-reinforced composite. It is logically sound to use the well-established finite element method (FEM) to analyse heterogeneous structures, but modelling tool penetration in a heterogeneous structure, for which adequate material separation criteria are required, is a complex issue. As various
types of machining operations are performed on bone, the tool’s cutting edge may cut along or across the osteons.

Modelling tool penetration in bone composite structure using FEA is complicated for three main reasons: (1) assigning the exact material properties to the constituents, (2) implementation of material separation criteria ahead of the tool in constituents at the interfaces of the constituents and (3) convergence issues in the presence of remeshing across the constituents’ boundaries. In addition, meshing all the details of constituent materials within the finite element model may be computationally challenging even to modern computers. It may also be necessary to remesh different zones with different criteria. Simplifying the fibre-reinforced composite as an equivalent homogeneous material (EHM) at the length scale larger than the characteristic length of a microstructure (e.g. a size of the representative volume element) for modelling purposes is a common practice. So the bone was approximated with the EHM in our simulations. The basis of this approximation and the procedure to obtain the effective elastic properties is explained in Section 5.4.

8.4 FE Model Setup

The MSC.Marc general FE code [182] was used in the development of a two-dimensional FE model of bone cutting. A 2 mm × 1 mm rectangular fragment of the bone was used in simulations. The cutting tool was modelled with rigid lines as the stiffness of tool material was much greater than that of the bone. The DOC was varied between 0.1 mm to 0.3 mm as discussed in Chapter 6. A “touching contact” was defined between the work piece and cutter; it started as the tool touched the work piece and continued until the end of simulations despite several remeshings. Figures 8.2 shows the components of the cutting

![Figure 8.2 Geometric representation of FE bone cutting model for CC](image)
system. To model the convection process, the convection heat transfer coefficient \((h)\) was applied to all the elements enclosed by the area ABCD as shown in Figure 8.3.

![Figure 8.3 Modelling convection in FE bone cutting model](image)

The FE model presupposes formation of a continuous chip, which was in good agreement with the results of our high-speed filming experiments of bone drilling and plane cutting (for DOC \(\leq 0.3\) mm) and other experimental studies on cutting of cortical bone [98][181]. All the simulations were continued until the stage of full chip formation was achieved, i.e. cutting forces, stresses and strains reached their saturated levels. The processes of CC and UAC in simulations were modelled from the moment of initial contact between the cutting tool and work piece at \(t=0\) to a state of the fully formed chip.

The updated Lagrangian formulation of FEA was used to model the cutting process. It allowed the element to change its shape with the material’s flow, and the calculation embedded a computational mesh in the material domain and solved the problem for the position of the mesh at discrete points in time. The updated Lagrangian formulation includes all kinematic nonlinear effects due to large displacements and rotations. The deformation method, using a plastic flow of the material under the action of the moving cutting tool as a separation criterion, was chosen for numerical simulations due to its better convergence properties (see also Sections 4.4 for the literature survey). The coupled thermomechanical analysis was performed with a staggered solution procedure. In that method, a heat transfer analysis was performed first, followed by a stress analysis at each time step. Experimentally, to get stable cutting forces and temperatures, it is necessary for the work piece and tool to be in contact for a reasonable amount of time. From our plane cutting experiments, it was observed that a time approximately of 4 seconds was required
to get the forces stable which was more than the simulation time. The main cutting force on the tool in the cutting process was calculated in the direction parallel to the cutting. The sequence with which the model was generated is shown in Figure 8.4.

8.4.1 Element Type

Element type 11 available in MSC.MARC element library was used in our simulations. It is a four-node, isoparametric, arbitrary quadrilateral designed for plane strain applications. As this element uses bilinear interpolation functions, the strains are constant within the element. This results in a poor representation of shear behaviour. The shear (or bending) characteristics can be improved by using alternative interpolation functions. The assumed strain procedure was flagged through the GEOMETRY option. This element is preferred over higher-order elements when used in a contact analysis. For nearly incompressible behavior, including plasticity or creep, it is advantageous to use an
alternative integration procedure. This constant dilatation method, which eliminates potential element locking, is flagged through the GEOMETRY option. The chosen type of elements must be used within the Updated Lagrange framework.

![Figure 8.5 Isoparametric quadrilateral plane strain element](image)

For a number of linear elements in Marc (Elements 3, 7, 11, 160, 161, 163, and 185), a modified interpolation scheme can be used, which improves the bending characteristics of the elements. This allows the ability to capture pure bending using a single element through the thickness. It was activated by use of the ASSUMED STRAIN parameter. This can substantially improve the accuracy of the solution though the stiffness assembly computational costs increases [182].

For each stress element, there is at least one corresponding thermal element, enabling the same mesh for both the heat transfer and thermal stress analyses. The associated heat transfer element in MSC MARC is Element type 39 (a four-node, isoparametric, arbitrary quadrilateral). Heat transfer elements are also employed to analyze coupled thermo-electrical (Joule heating) problems. As this element uses bilinear interpolation functions, the thermal gradients are constant within the element. The conductivity of this element is formed using four-point Gaussian integration.

A reduced integration scheme for these elements was used to determine their stiffness matrix. In such a reduced scheme, the integration is not exact; the contribution of the highest order terms in the deformation field is neglected. Reduced integration elements have specific advantages and disadvantages. The most obvious advantage is the reduced cost for element assembly. Another advantage is the improved accuracy which can be obtained with reduced integration elements for higher-order elements. The increase in accuracy is due to the fact that the higher-order deformation terms are coupled to the lower order terms. The coupling is strong if the elements are distorted or the material’s compressibility becomes low. The higher-order terms cause strain gradients within the element, which are not present in the exact solution. Hence, the stiffness is overestimated. Since the reduced integration scheme does not take the higher-order terms into account, this effect is not present in the reduced integration elements [182]. MARC uses a reduced
integration scheme to evaluate the stiffness matrix or the thermal conductivity matrix for a number of isoparametric elements. The mass matrix and the specific heat matrix of the element are always fully integrated. For lower-order, 4-node quadrilateral elements, the number of numerical integration points is reduced from 4 to 1.

8.4.2 Mechanical boundary conditions

It was not feasible to apply boundary conditions (constraints) directly to the work piece due to remeshing necessary to account for high deformations in the chip. To resolve that, rigid lines AD, DC and CE (see Figure 8.2) were created at the left, bottom and right edges of the work piece and were permanently “glued” to it. Mechanical boundary conditions were applied at those lines, which were transferred to the work piece elements. Recent developments in the software now allow the user to apply boundary conditions directly to the remeshing parts, but the results may not be accurate, as sometimes the values of the boundary condition are changed due to calculation errors. Our initial numerical simulations confirmed the occurring of such errors.

Kinematic boundary conditions were applied to the modelled work piece and cutting tool. Vibration of the tool provide separation of the tool from the chip at each cycle of ultrasonic vibration, thus transforming the constant contact at the tool–chip interface into an intermittent one. In CC, the lines were given the velocity in the positive x-direction and the tool was held stationary/fixed in all degrees of freedom. In UAC, the work piece was given the prescribed velocity in the positive x-direction as shown in Figure 8.6 and the tool was vibrated harmonically about an equilibrium position. The FE model with the prescribed boundary conditions is shown in Figure 8.6. They represented the constraints similar to our plane cutting experiments, where a cutting tool was fixed and the bone specimen, held in a fixture, was given the velocity against the cutting tool. Mathematical formulation of the boundary conditions is as follows:

\[
V_x|_{AD}, V_x|_{DC}, V_x|_{ICE} = V_c, \\
V_y|_{AD} = 0, V_y|_{DC} = 0, V_y|_{ICE} = 0, \\
V_z|_{AD} = 0, V_z|_{DC} = 0, V_z|_{ICE} = 0, \\
\]

Where \(V_c\) is the cutting speed. The cutting tool was modelled as a rigid stationary body (in simulations of the CC process) or vibrated harmonically in the cutting direction in UAC simulations:

\[
u_x = -a \cos \omega t, \quad u_y = 0,\]
where $\omega = 2\pi f$, $f$ is the frequency of vibration (kHz) and $a$ is amplitude ($\mu$m).

Figures 8.6 Kinematic boundary conditions on the FE meshed work piece. (a) CC: tool is fixed and work piece moves with cutting velocity in the positive x-direction. (b) UAC: work piece moves with cutting velocity in the positive x-direction and the tool oscillates along the axis about a fixed mean position.

8.4.3 Work–tool interaction

Friction between the cutting tool and the work piece and the work of plastic deformation during a cutting process are the main sources of heat generation resulting in a temperature rise in the tool-work piece interaction zone. For true estimation of such changes it is important to perform thermomechanically coupled simulations. In real cutting processes, due to high stresses, high strain rates and high temperatures, a high level of mechanical power is dissipated in the tool–chip interface thus leading to many structural modifications of the contacting pieces. Therefore Shih [183] showed that there is no universal contact law that can predict friction forces for a wide range of cutting conditions. The friction character at the tool–chip interface is difficult to determine since it is
influenced by many factors such as the cutting speed, contact pressure and temperature. In previous computational studies, the frictional contact at the tool–chip interface in cutting was considered mainly in terms of the two friction models which were employed for this purpose. They include the Coulomb friction model, with friction stress being proportional to normal pressure at the interface \( \tau = \mu \sigma_n \) \[125][184] and the shear friction model \( \tau = \mu k \), where \( k \) is shear yield strength) \[111][184].

The classical Coulomb model is unable to adequately reflect friction processes due to generation of high contact stresses at the tool–chip interface leading to significant friction forces. Hence, the shear friction model [182] was used in simulations. Here, the friction force depends on the fraction of the equivalent stress of the material and not the normal force as in the Coulomb model. Thus, friction stress was introduced in the following form:

\[
\sigma_{\text{fr}} \leq -\mu \frac{\bar{\sigma}}{\sqrt{3} \pi} \left( \text{sgn}(\nu_r) \arctan \left( \frac{\nu_r}{\nu_{cr}} \right) \right)
\] (8.2)

where \( \bar{\sigma} \) is the equivalent stress, \( \nu_r \) is a relative sliding velocity, \( \nu_{cr} \) is a critical sliding velocity below which sticking occurs, \( \mu \) is a friction coefficient.

### 8.4.4 Modelling heat transfer

During the mechanical part of the analysis, heat generated by plastic deformation and friction was calculated. In the finite element model, heat generation due to plastic deformation and friction at the tool–chip interface was modelled as a volume heat flux which included both plastic dissipation and frictional heating. Heat convection was assumed to be the primary mode of heat transfer, which occurred between the work piece material and the environment. Thermal processes in cutting comprises:

- plastic heat generation in the work piece material due and frictional heating at the tool-chip interface,
- contact heat transfer between the work piece and the tool,
- convective heat transfer from free surfaces of the work piece to the environment given by:

\[
-k \frac{\partial T}{\partial n} = h(T - T_o),
\] (8.3)
where \( n \) is the outward normal to the surface, \( K \) is thermal conductivity and \( h \) is the convective heat transfer coefficient at the material environment interface and \( T_a \) is the ambient temperature. The thermal flux, passing from the chip to the cutter at the contact length, is described as follows:

\[
q = H(T_b - T_t) \tag{8.4}
\]

where \( H \) is a contact heat transfer coefficient, \( T_b \) and \( T_t \) are the bone and tool surface temperatures, respectively. The measurement of contact heat transfer is a complex matter and depends on a number of cutting parameters such as the cutting speed, work and tool material properties, contact pressure at the interface and the nature of the surfaces of the contact bodies. As the exact value of the contact heat transfer coefficient between the tool and work piece was not known, no heat transfer to the cutting tool was modelled in those simulations.

In actual practice to cool the orthopaedic tool and minimize the temperature rise in the bone, a physiological solution is applied as a coolant (irrigation). In orthopaedic bone cutting, the bone is traditionally irrigated by a saline solution as a coolant to maintain the temperature at a controlled level. The value of \( h \) was provided in the simulations to represent convection (irrigation). The cooling environment was modelled by applying a convective heat transfer coefficient \( (h) \) of the physiological saline solution commonly used in bone surgeries (see Figure 8.3). In this study, the convective heat transfer coefficient of a saline solution as measured by Tangwongsan [185] for an average flow rate of 1000 ml/min, was used in simulations. The initial temperature \( (37^\circ C) \) was applied as nodal quantities to the work piece mesh. Mathematically the initial thermal conditions used in simulations are as follows:

\[
T_b|_{t=0} = T_a ;
\]

where \( T_b \) is the initial temperature of the bone and \( T_a \) is the ambient temperature \( (20^\circ C) \).

### 8.4.5 Mesh zoning

Cutting is a process in which a work material is severely deformed when the chip is separated from the cutting zone. During the simulations, elements in the process zone could become highly distorted, leading to degradation in solution accuracy and premature termination of analysis. A high mesh density at the cutting zone is necessary to avoid convergence issues and to get reliable calculations of cutting parameters (forces, stresses, temperatures). This, however, should be compromised as the increased number of elements
is computationally expensive. In MSC MARC, remeshing can only be carried out when a portion of the FE model is defined as a contact body. Therefore, contact bodies are expected to be used in the analysis with global remeshing [182]. The basic steps, which are automatic in global remeshing, are as follows:

1. Analysis performs remeshing based on the defined remeshing criteria.
2. A new mesh of the deformed shape of the contact body is generated and is called alone mesher or an internal mesher.
3. The new mesh is checked for penetration to avoid contact loss to other contact bodies.
4. The data mapping is performed that transfer the data from the old deformed mesh to the new mesh.
5. The contact tolerance is measured automatically and the contact conditions are re-established.
6. After a new mesh is created, the new contact is detected using the new mesh contact boundary conditions and are automatically updated.
7. The analysis completes global remeshing and continues based on the newly generated mesh.

The process of remeshing is shown in Figure 8.7.

Simulations were run to check for better convergence and smooth contours. Varying mesh densities were tried to minimize the total number of elements and nodes for computational efficiency. The simulation interval was subdivided into 2,000 time increments of $1.25 \times 10^{-6} \text{s}$. From our initial simulations, the analysis was found to diverge around 7–10 increments depending on other cutting parameters. So, remeshing was forced after every 5 increments in the work piece and chip volumes, providing the new
mesh to generate. Analyses were converged using the element edge length of 0.015 mm. To see the effect of mesh density on the numerical results, the element edge length was varied from 0.005 mm to 0.015 mm. Since the difference in the calculated stress, force and temperatures, for the range of mesh density used was below 4%, elements of larger edge length (0.015 mm) were used in simulations for computational efficiency. The total number of elements in the work piece before remeshing were 800 and increased to a total number of approximately 17000 when the first remeshing step was applied. As the mesh density was controlled by a fixed value of element edge length, the number of elements were almost the same in the subsequent remeshing steps. To check the contact stability in the iteration process, the motion of the node was automatically controlled with regard to its penetration of a surface by a certain amount.

8.5 Material Properties

To obtain reliable results from Finite Element (FE) simulation of machining processes, it is necessary to have as input the properties of the work piece and tool materials as well as the characteristics of the tool/chip interface. These input parameters include physical and thermal data, friction and heat transfer, and, most importantly, the flow stress of the work piece material under high strain, strain rate and temperature conditions that exist during the process. Certain mechanical properties of the bone tissue, particularly its modulus and ultimate strength, have been found to vary at different strain rates.

The Johnson-Cook (JC) material model was proposed for cutting of cortical bone, which accounts for strain-rate and temperature effects. The material was considered as an elastic-plastic with bilinear strain hardening. The JC material model was widely used in metal cutting simulations [111][112] and it is given by

\[
\sigma_y = \left( A + B \varepsilon_p^a \right) \left( 1 + C \ln \left( \frac{\dot{\varepsilon}_p}{\dot{\varepsilon}_o} \right) \right) \left( 1 - T^* \right),
\]

where \( A \), \( B \), \( C \) and \( n \) are constants and \( \sigma_y \), \( \varepsilon_p \), \( \dot{\varepsilon}_p \) and \( \dot{\varepsilon}_o \) are the yield stress, plastic strain, effective plastic strain rate and reference strain rate, respectively. \( T^* = \frac{T - T_{room}}{T_{melt} - T_{room}} \), where \( T_{room} \) and \( T_{melt} \) are the room and melting temperatures, respectively. Term \( T^* \) was omitted in the simulations at this stage of investigations due to unavailability of reliable data and due to small temperature variations calculated in the material about 80°C, with no significant thermal effects expected in this range. In our bone cutting experiments (plane cutting and drilling), the chip generated was continuous as in
metal cutting (see section 9.5.1 and section 6.5.3). The similarity in the chip formation process in bone cutting and metal cutting together with similar strain rate sensitivity justified the use of the JC model for FE model of cutting bone. The parameters in the JC material model are sensitive to the computational algorithm used to calculate them. The procedures were illustrated for data analysis of the bone tension tests. For that study, the response at different strain rates was emphasized.

Based on our experimental results, it was assumed that the effect of strain rates higher than 1 s\(^{-1}\) can be neglected (see Section 5.2.4). That provided an opportunity to utilize the Johnson-Cook material model, incorporating strain rate and temperature effects in metals, for cutting of bone. The JC constants were calculated by fitting the equation into the stress-strain graphs obtained from the tension tests on the bone as shown in Figure 8.8; their magnitudes used in simulations are given in Table 8.2. The material properties of cortical bone obtained from the tests and thermal properties from the literature are provided in Table 8.3.

![Figure 8.8](image)

**Figure 8.8** Experimental stress-strain data for tension and their approximation with the Johnson-Cook material model

**Table 8.2** Johnson-Cook parameters used in cutting simulations

<table>
<thead>
<tr>
<th>A (MPa)</th>
<th>B (MPa)</th>
<th>C</th>
<th>n</th>
</tr>
</thead>
<tbody>
<tr>
<td>50</td>
<td>101</td>
<td>0.03</td>
<td>0.08</td>
</tr>
</tbody>
</table>
To study the effect of different processing parameters on cutting forces, temperatures and the effect of different variables on these process parameters, a series of simulations were performed. The results of these simulations are presented in the next chapter.

### 8.6 Limitations of Finite Element Model

The FE developed FE model is of value but has some limitations as listed below.

1. The current model is based on a Johnson-Cook material model which is suitable for a strain-rate sensitive material and can include temperature effects. In the model, temperature effects were not considered due to the lack of such data for cortical bone tissue.

2. Material properties used were derived from tension tests performed in the longitudinal direction of the bone. The cutting action of the tool edge was considered as separation of the chip from the work piece in tension.

3. The maximum strain rate applied to the specimens was limited to 1s⁻¹ due to limitations in specimen dimensions. It was assumed that the stress-strain curve was not affected by the strain rate above 1s⁻¹.

4. The cutting action of the drill was idealised as two-dimensional at the outer corner of the drill cutting lip. The cutting of the drill was considered to take place at the extreme corner of the drill where the cutting speed is maximum.

5. The value of friction coefficient, 0.38, was taken from the literature reported for a similar wood cutting process.

6. There is a wide range of thermal properties for bone reported in the literature. This model relied on the thermal conductivity and specific heat values taken from the literature. The model can be improved by measuring these properties for the cortical bone used in cutting experiments.

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Effective elastic modulus (GPa)</td>
<td>20.0</td>
<td>Calculated (Section 5.4)</td>
</tr>
<tr>
<td>Thermal conductivity (W/m K)</td>
<td>0.56</td>
<td>[53]</td>
</tr>
<tr>
<td>Heat capacity (J/m³K)</td>
<td>2.86×10⁶</td>
<td>[46]</td>
</tr>
</tbody>
</table>
Chapter 9. Finite Element Analysis of Bone Cutting

9.1 Introduction

The development of thermomechanically coupled FE models for bone cutting was presented in the previous chapter. This chapter presents the results obtained by numerical simulations based on the model. The processes in UAC were simulated for 60 cycles of vibration (i.e. a cutting time of 3 ms), the same period was modelled for conventional cutting. It covers the period from the initial contact of the cutting tool with the bone to formation of the fully shaped chip. This simulation time was sufficient for the cutting process to reach the steady state cutting process with saturation of the level of force, stress and temperatures in the work piece material. Each simulation took about 5 hours of calculation time on the Pentium IV, 2.8 GHz workstation with 2048 MB RAM. Both the CC and UAC simulations were divided into the same number of increments for better comparability of the results. Thermo-mechanical coupling was achieved by calculating the nodal temperatures followed by the deformation and frictional effects in the work piece material at each solution step. A series of different simulations were conducted to study the effect of process parameters on level of stresses, cutting forces and temperature at the cutting zone. Table 8.1 shows the details of variables of the cutting process. These include the cutting speed, friction, depth of cut (DOC), amplitude and frequency of vibration in UAC.

In order to have a clear understanding of the features of the process under conditions of superimposed ultrasonic vibration, a set of simulations was performed to calculate the influence of various parameters of UAC. For those simulations all the variables common to CC were kept constant and only the frequency and amplitude were varied. The following sections present the results obtained by means of FE simulations with MARC.MENTAT.

9.2 Calculations of Stresses

In bone cutting, two types of stresses are of importance: (1) stresses generated during the chip formation which affect cutting process and (2) residual stresses that affect the bone-implant interface strength. The current work was aimed at investigating the levels of stresses induced in bone, caused by a cutting tool penetrating into it, in conventional or ultrasonically-assisted modes and their comparison. The effect of ultrasonic frequency and amplitude on material stresses is discussed. The developed FE model is fully thermo-
mechanically coupled in order to properly reflect interconnection between the thermal and mechanical processes accompanying cutting of bone.

### 9.2.1 Stresses in bone

One of the important mechanical factors affecting bone remodelling is the stress in bone tissue. For remodelling to occur, bone should possess some stress present in its tissue to stimulate the osteocyte over a relatively long period. This stress will be a type of residual stress that satisfies the force equilibrium condition in the bone material [186]. During cutting of bone, these stress conditions will change which will necessitate their measurements for analytical and computational models of bone remodelling and enhance understanding of the mechanics of the bone-screw interface strength.

In our simulations the stress distribution in the bone material changed with the penetration depth and achieved a stable value exceeding the material yield limit when the chip was formed. The increase in stress levels beyond the yield strength of the material was due to strain hardening. The separation of the material in the cutting process occurred in the vicinity of the cutting edge that moved along the prescribed straight line in the simulations. The material failed when the stress in it exceeded the critical value. Simulations were performed with various cutting speeds as given in Table 8.1 of chapter 8. A contact was defined between the tool and the work piece, where relative motion between them was allowed under friction conditions. The software [182] used in simulations has the capability to detect the contact automatically.

In CC, the stress reached almost a constant value of 140 MPa at 0.7 ms and remained stable for the remaining cut (up to full chip formation). At the initial stage of cutting, stresses were relatively small and concentrated around the tool tip in both the chip and machined surface. As the cutting process continued until the shear band was fully formed, the deformation increased and spread over a zone where shear banding takes place. The average calculated shear stress was 45 MPa. The deformation zones produced in the material when the tool advanced are shown in Figure 9.1; they are characterised by high levels of stresses (bright colours in the Figure 9.1). It was noticed that the field of maximum equivalent stress was almost the same in primary and secondary deformation zones. The maximum equivalent stress fluctuated between 135–140 MPa and the measured shear angle was approximately 40°. The stress fluctuation can be attributed to numerical errors inherent to finite element simulations as well as to the frequent remeshing scheme. The shape of chips created while drilling into bone was analysed microscopically by Jacobs [86]. The drilling speed and feed rate used in he experiments were respectively
from 100 to 2360 rpm and 4.23 to 21.2 mm/s. The analysis showed that the majority of chips recovered indicated shear as the mechanism of failure.

Figure 9.1 Deformation zones produced when tool penetrated into material in conventional cutting of bone (cutting speed, $V_c = 300$ mm/s)

9.2.2 Effects of ultrasonic vibration on stresses

The level of stresses induced by the tool penetration in UAC were compared to those of CC for identical cutting conditions. To fully understand the material’s response in the interaction with the ultrasonic tool, the stress variation with tool motion was split into two stages: (a) when the tool penetrated into the material in the forward direction and (b) when the tool travelled away from the chip. In UAC, during the continuous chip formation, a dynamically changing stress distribution in the process zone was observed. Figure 9.2 shows characteristic stages of a single cycle of vibration that took about 0.05 ms. When the tool moved forward, the stress increased with tool penetration and attained a maximum value of approximately 140 MPa near the tip of the tool. The attainment of the maximum penetration depth was characterized by the highest levels of generated stresses in the process zone and indicated the end of loading/penetration stage (Figure 9.2a). When the tool moved in the opposite direction, the level of maximum stress dropped to an average value of 30 MPa (Figure 9.2b). Those were the residual stresses induced in the material due to its plastic deformation. During the penetration stage, the cutting force increased in the similar fashion and achieved a peak value at maximum penetration depth (end of loading). The maximum stress values were almost the same for all loading/cutting stages in UAC. As these maximum stresses were reached only during the loading part of the ultrasonic cycle, the average over a cycle stress in the bone material was considerably smaller than that in CC. For maximum penetration of the tool in the ultrasonic cycle, the
level of stress induced in the material by the tool below its cutting tip (i.e. below the cut surface) decreased rapidly with increasing distance from the cut surface (Figure 9.3).

![Figure 9.2](image.png)  
**Figure 9.2** Distribution of von Mises (equivalent) stresses in MPa for a single cycle of ultrasonic vibration: (a) tool penetration; (b) tool moving away ($f = 20$ kHz and $a = 20 \mu$m).

![Figure 9.3](image.png)  
**Figure 9.3** Variation of material stress with distance from the cut surface at maximum tool penetration. $f = 20$ kHz and $a = 20 \mu$m.

To study the effect of amplitude of ultrasonic vibration on the stress developed in the bone material, the amplitude was varied between 5 and 30 $\mu$m (see Table 8.1) while other parameters of UAC were kept constant. It was found that the amplitude of ultrasonic vibration did not influence the maximum stress magnitude since the process of cutting was simulated for the stress strain relation where a maximum stress limit was provided as an input for the cutting process. The maximum stress was achieved at a penetration depth of
about 10 µm (for \( f = 20 \text{ kHz} \) and \( a = 20 \mu m \)) and remained nearly constant until the end of the loading stage of the cycle. For a constant frequency, the only effect of increasing amplitude was the increase in duration, for which the stress induced in the material retained its maximum magnitude. The change in ultrasonic frequency resulted in the varying speed, at which the stress state changed in the material. The variation in the cutting speed within the studied interval (Table 8.1) was found to have no influence on peak stresses.

9.3 Analysis of Cutting Forces

One of the objectives of this study is the assessment of the levels of cutting forces for a given set of cutting parameters. Low-level fluctuations of the cutting force were observed in our simulations; they can be attributed to the remeshing as well as numerical errors inherent to FE simulations (Figure 9.4). The fluctuation in the force is due to the mapping of the stress and strain data on the new mesh generated in remeshing process. This level of fluctuations can be reduced by using the finer meshes in the tool-work piece interaction zone and also by increasing the number of time increments (of a smaller magnitude) and introducing more remeshing cycles between increments. A denser mesh will, however, require considerable computational power and a large memory. The force at peak value was therefore smoothed when presenting graphically the obtained results below in order to eliminate those low-level fluctuations.

The kinematic boundary conditions applied to simulate the level of cutting forces in

![Graph](image-url)

**Figure 9.4** Evolution of cutting forces in CC (fluctuations are due to remeshing). DOC = 0.2 mm, \( \psi = 10 \mu m \), cutting speed \( V_c = 200 \text{ mm/s} \)
CC and UAC are described in Section 8.4.2. The software has the capability to combine all the forces on the tool and resolve to a single point, normally a centroid of the body [182]. The FE model was validated by comparing the calculated results for the cutting force with those obtained from experiments of plane cutting of bone (see Section 9.5.2). In all those simulations the forces developed rapidly (within 0.05 ms) and achieved saturated levels at approx. 0.2 ms; that level remained nearly constant in simulations for the remaining length of cut. Several case studies were implemented to analyse the effect of various parameters on the level of cutting forces. Their results are presented below.

### 9.3.1 Forces in CC

The fully parametric FE model allowed to change cutting parameters and calculate their effect on the magnitude of the cutting force. The obtained results of simulations showed a significant difference in forces acting on the cutting tool when the DOC was varied. The magnitudes of the longitudinal cutting force for various levels of DOC are shown in Figure 9.5. An increase of 66% in the cutting force was observed when the DOC was increased from 0.1 mm to 0.2 mm. Another 62% increase resulted from the DOC change from 0.2 mm to 0.3 mm. The increase in cutting force was due to the fact that the tool had to deform a thicker layer of bone in the process zone at the higher DOC with increased resistance. The increase in the amount of friction (due to larger tool-work contact) was another contributing factor.

![Figure 9.5](image.png)

**Figure 9.5** Effect of DOC on cutting force. \( \psi = 10 \, \mu m, \alpha = 20^\circ, V_c = 200 \, \text{mm/s} \)
To account for the change in the cutting force with an increase in the material removal rate, the tool speed was varied from 100 mm/s to 600 mm/s. The cutting forces remained almost unchanged for the entire range of rates. The cutting force was also observed to be unaffected by the range of velocities used in plane cutting experiments (see Section 6.3.4). Those results demonstrate that thickness of the removed layer is the parameter defining the cutting forces for a studied range of cutting speeds. The results obtained in drilling experiments demonstrated another type of response which showed a decrease in thrust force with increase in drilling speed.

The exact definition of frictional interaction between bone and cutter is not reported in the literature. A significant difference was observed in the level of forces on the cutting tool when the friction coefficient was increased from $\mu = 0$, i.e. frictionless conditions, to lubricated cutting, with $\mu = 0.35$. The magnitude of cutting forces obtained in simulations with and without friction are shown in Figure 9.6. An increase of 40% was observed with introduction of friction for DOC = 0.3 mm, whereas at the lower DOC (0.1 mm) an increase of 58% was noted.

![Figure 9.6](image)

**Figure 9.6** Increase in cutting force with introduction of friction at tool/work piece interface. $\psi = 10$ µm, $\alpha = 20^\circ$, DOC = 0.3 mm, $V_c = 200$ mm/s

The tool nose radius $\psi$ was changed from 0 to 15 µm in our simulations. A comparison of forces acting on the cutting tool is shown in Figure 9.7. The force increased by a considerable amount with an increase in the nose radius. The cutting force increased by 37% when tool nose radius was increased from 0 to 15 µm. A sharp edge of the tool caused stress concentration in work piece elements hence enabling the tool to penetrate with less energy. In the case of a larger edge radius, more elements came into contact with the curved edge and higher energy was required for penetration.
9.3.2 Analysis of forces in UAC

In simulations of UAC, the work piece was moved with a constant velocity in the X-direction, and the tool was vibrated with the required frequency and amplitude. Simulations were conducted using models of UAC and CC to compare the forces acting on the cutting tool during cutting. The force pattern obtained when ultrasonic vibration was applied to the tool is shown in Figure 9.8. Since the cutting tool was in permanent contact with the work piece during CC simulations, a practically non-changing force was present in the cutting tool, while in UAC in a single cycle of vibration from the moment of the first contact with the chip, the forces start increasing with penetration and attained levels.
somewhat higher than the average force predicted in CC when the tool reached the maximum penetration depth (Figure 9.9). The force dropped as the tool started moving away from the work piece and eventually vanished when separated from it. The forces stayed close to zero level until the cutter came again into contact with the chip in the next cycle of ultrasonic vibration.

Figure 9.9 Comparison of calculated forces in cutting tool in simulations of CC and UAC \( \psi = 10 \, \mu m, \text{DOC} = 0.3 \, mm, V_c = 200 \, mm/s \). For UAC, \( f = 20 \, kHz \); and \( a = 20 \, \mu m \)

Simulations showed a consistent force value in the cutting tool for CC. In case of UAC, the maximum value of this force was almost equal to that in CC at the peak. Since the force is changing from a zero level to a maximum and then dropping to zero in each vibration cycle of UAC, the average value of force over the time of a force cycle was considered in calculations. For quantitative analysis, averaging of the two values for one full cycle of vibration (Figure 9.9) was performed using the following relation:

\[
\langle F_c \rangle = f \int_{t_s}^{t_c} F_c \, dt,
\]

where \( f \) is the frequency, \( t_s \) and \( t_c \) are the start and end times of the vibration cycle.

Averaging of the values for UAC shows that the forces in the case of UAC are lower than the forces in the case of CC. Comparison of the obtained values with the experimental results showed a good agreement (see Section 9.5.2).

A separate study was conducted to analyze the effect of frequency and amplitude of ultrasonic vibration on the forces in the cutting tool. To investigate the effect of ultrasonic
frequency on cutting force, the DOC, amplitude and cutting velocity were kept constant. The force was observed to drop linearly with increasing frequency as shown in Figure 9.10. The averaged force dropped by 62% when frequency was increased from 5 to 30 kHz. In UAC, the tool and the work piece kept separating and contacting producing a pulsating cutting force. Increasing ultrasonic frequency (for other cutting parameters being constant) decreased the total amount of contact time between the cutting tool and work piece thus reducing the averaged level of cutting force. The drop in the force can also be explained by the decrease in the area under the force-time curve for increased frequency. The effect of ultrasonic frequency on cutting force was confirmed by measurements in plane cutting experiments for lower cutting speeds.

![Figure 9.10](image)  
**Figure 9.10** Effect of ultrasonic frequency on forces in cutting tool. DOC = 0.2 mm, a = 10 µm, ψ = 10 µm, $V_c$ = 200 mm/s

FE simulations were also conducted to study the effect of vibration amplitude on the forces acting on the cutting tool in UAC. Figure 9.11 shows the influence of ultrasonic amplitude on cutting force. A slight increase in the peak force with increase in the amplitude was observed. However, the averaged cutting force decreases with an increase in the amplitude; a decrease of approx. 23% was recorded for an increase from 5 µm to 15 µm and a further approx. 22% decrease was observed when the amplitude was increased from 15 µm to 30 µm. The drop in the force was due to the shape of the chip and change in the shear angle. The effect of amplitude on cutting force was similar to that produced by frequency.
9.4 Analysis of Cutting Temperatures

The heat generation in cutting processes strongly depends on physical and chemical properties of the material, cutting conditions and the cutting tool geometry. To investigate the influence of cutting parameters (see Table 8.1) on bone temperatures in the cutting zone, a series of simulations was performed. Temperature values were calculated with the developed model and compared for cutting conditions with and without irrigation. The influence of bone’s thermal properties on the depth of thermal necrosis was studied. Low-level fluctuations were found in temperature values in the chip due to remeshing and were smoothed in the results. The temperature rise was due to the irreversible deformation in the process zone as well as to the friction between the tool and the work piece. After reaching the saturated level when the chip was fully generated and the tool-chip contact length reached a constant value, the temperature stayed constant for the remaining length of cut.

The regions where heat was generated during the cutting process are shown in Figure 9.12. They are: (A) the primary deformation zone where plastic deformation takes place in the shear plane, (B) the secondary deformation zone where additional deformation and sliding friction occurs at the tool/chip interface contact, and (C) tertiary deformation zone, which is produced due to the rubbing contact between the tool flank face and the new machined surface of the work piece. The calculated temperature distribution at various regions for the condition when the chip was fully developed is shown in Figure 9.13. The highest proportion of heat was produced in the chip (in the secondary deformation zone, i.e. the front face of the tool); however, the chip was considered to be no more a part of the
bone and its temperature was ignored. Instead, the temperature produced in the work piece material near the flank of the tool (see zone indicated by an arrow in Figure 9.13) was considered for the analysis. From preliminary simulations performed, no significant effect of tool nose radius on cutting temperature was found; so it was kept constant (ψ = 10 µm) in all simulations pertinent to temperature analysis.

![Figure 9.12](image1.png)

**Figure 9.12** Heat zones in orthogonal cutting: (A) primary deformation zone, (B) secondary deformation zone, and (C) tertiary deformation zone

The value on the ordinate (ΔT) of the subsequent plots, represents the rise in temperature in bone material above 37°C (ΔT = 33°C). The simulation interval, remeshing frequency and analysis time were kept similar to those used in analysis of stresses and forces. The simulated cutting temperatures were compared with experimental results obtained in bone drilling tests and are described in Section 9.5.3.

![Figure 9.13](image2.png)

**Figure 9.13** Temperature distribution in cutting region: (a) close-up view of cutting edge – arrow indicates temperature produced below freshly cut surface; (b) work piece-tool interaction zone (Vc = 400 mm/s, DOC = 0.2 mm, h = 0)
9.4.1 Temperatures vs. tool speed

The effect of cutting speed on the temperature rise in the bone material was investigated based on the developed FE model. With tool penetration into the material, the chip starts separating and sliding along the tool rake. Higher velocities cause a quicker development of the chip and, hence, the temperature increases faster too. The effect of tool velocity on the temperature calculated in the work piece material adjacent to the tip of the tool, i.e. the area of maximum temperature, without cooling environment is shown in Figure 9.14. A similar influence of the cutting speed on temperatures in bone was briefly discussed by Hillery [67]. A temperature rise with increase in the cutting velocity was due to the increase in the rate of irreversible deformation. For the material properties used in the model, the thermal necrosis threshold level was registered at the cutting speeds of 400 mm/s and 600 mm/s while the temperature level achieved at 200 mm/s was well below it. The cutting speed of 200 mm/sec corresponds to the linear velocity at the outer periphery of a 4 mm diameter drill rotating at approximately 1000 rpm.

![Figure 9.14 Effect of cutting speed on temperature evolution in bone (DOC = 0.2 mm, \( h = 0 \), bone’s initial temperature 37°C) (NTL – necrosis threshold level)](image)

9.4.2 Effect of cooling

In orthopaedic bone cutting, the bone is traditionally irrigated by a saline solution used as a coolant to maintain the temperature at a controlled level. The effect of cooling on the maximum temperature in bone was studied in our simulations and, not surprisingly, a significant decrease in temperature was achieved when the irrigation environment was employed as shown in Figure 9.15. Without cooling, thermal necrosis was found for the
range of cutting speeds above 300 mm/s (for DOC = 0.2 mm). In comparison, when the cooling condition was applied, the calculated rise in temperature did not reach the necrotic level for the cutting speeds below 550 mm/s.

![Figure 9.15](image)  
**Figure 9.15** Effect of cutting speed on temperature with cooling ($h = 8000 \text{ W/m}^2\text{K}$) and without it ($h = 0$), DOC = 0.2 mm

### 9.4.3 Effect of depth of cut

To investigate the effect of DOC on bone temperature, simulations were performed for a constant cutting speed of 400 mm/s. The cutting speed was arbitrarily chosen and was kept constant to calculate the effect caused by DOC only. It is evident from the results presented in Figure 9.16 that the temperature increased non-linearly with increasing DOC; the slope of the curve slowly decreased with growing DOC. This can be related to the increase in frictional effects (increase in chip-tool contact length along the rake face) for a higher DOC. For the material properties used in simulations, the necrosis threshold level (NTL) was exceeded for DOC above 0.125 mm when no cooling was applied. Increasing the DOC above 0.2 mm did not result in a significant temperature growth in the work piece since the temperature increase was localised predominantly in the secondary deformation zone (labelled B in Figure 9.12) due to the increased contact length between the tool and the chip.
9.4.4 Investigating depth of thermal necrosis

To our knowledge, studies describing the depth from the cut site, at which necrosis in the bone tissue may occur, are limited in the amount of information. The maximum temperature in the bone material was found in the chip region adjacent to the tool rake near its flank. Heat generated beneath the cutting edge caused the temperature to rise below the freshly cut surface near the tool flank (see Figure 9.17). The term DTN (depth of thermal necrosis) refers to the depth (in micrometers) below the cut surface where necrosis level was reached. The axis of ordinates in Figure 9.17 represents the depth of the layer
measured from the cut surface in which the threshold level 70°C (ΔT = 33°C) was attained. The necrotic levels were taken when the chip was fully developed and temperature values stabilised in numerical simulations. The depth level was observed to depend strongly on the tool velocity and DOC, increasing with an increase in DOC. That was due to the fact that growing DOC amplified the frictional effect due to larger contact between the tool and work piece, resulting in a higher temperature rise in the bone material. The DTN was negligible up to the cutting speed of 200 mm/s for all values of DOC.

9.4.5 Effect of thermal properties of bone on Necrosis depth

Thermal properties of the bone material, which influence the resulting cutting temperatures, include its thermal conductivity and heat capacity. A temperature rise in the machined material generally decreases as these parameters increase. Bone thermal conductivity and heat capacity vary significantly among individuals [53], and it is important to investigate their effect on DTN. FE simulations were performed in which the thermal conductivity $K_b$ and heat capacity $C_b$ were varied by between -20% and +20% each from their nominal values given in Table 8.3. The ranges of thermal conductivity (0.45–0.67 W/m K) and heat capacity ($2.29\times10^6 – 3.43\times10^6$ J/m$^3$K) were used in simulations. According to the simulation results (Figure 9.18), thermal conductivity had a negligible effect on DTN, while a 20% increase in heat capacity resulted in a 46% decrease in necrosis penetration. A similar effect of the described properties on the necrosis propagation in bone drilling was reported by Davidson [75].

![Figure 9.18](image)

**Figure 9.18** Influence of change in bone thermal conductivity ($K_b$) and heat capacity ($C_b$) on depth of thermal necrosis ($V_c = 400$ mm/s, DOC = 0.2 mm)
9.5 Comparison of Experimental and FE Results

The previous section discussed the results of the finite element model of the bone cutting process. The results obtained from FE simulations were based on the mechanical properties of cortical bone obtained from experiments and thermal properties taken from literature. For a finite element model to be validated, the results are required to be compared either with experimental results or an analytical model of the process. This is not always possible either due to the complex nature of the process itself or the behaviour of the material in relation with the equipment for obtaining the required data. The measurement of stresses in the bone cutting process is a difficult task due to the use of strain gauges on bone surface near the regime of high deformations or delicate instrumentation in a non contact method such as spectroscopy. Therefore, comparison of stresses induced during the bone cutting process using experimental and FE techniques were not made at this stage of investigation. The next section compares the chip formation, force and temperature obtained from finite element simulations with experimental results of plane cutting and drilling.

9.5.1 Chip formation

As the cutting force and temperature measurements are strongly influenced by the chip formation mechanism in all cutting processes, chip shape was compared for experimental

![Figure 9.19 Chip formation in bone cutting process: (a) experimental plane cutting; (b) prediction of finite element model (ψ = 10 µm, α = 20°, V_c = 10 mm/s, DOC = 0.2 mm)](image)
and FE analysis. It was interesting to see that the chip formation in experimental plane cutting was no more continuous for a DOC larger than 0.3 mm. Therefore, the chip shape observed in experiments and those produced in numerical simulations, were compared for a value of DOC up to 0.3 mm. The kinematics of the tool-work piece interaction was kept similar in experiments and simulations where the tool was fixed and the work piece was moved with the prescribed speeds. To compare chip formation process shown in Figure 9.19, the tool rake angle, nose radius and cutting speed were kept the same in experiments and simulations. The chip generation process in experiments was observed using a digital camera focused on the cutting location on the outside surface of cortical bone. The cutting was along the longitudinal direction of the bone. The comparison of chip shape was for cutting speeds up to 16 mm/s which was well below the speed used in real bone cutting in orthopaedics. This necessitates more experiments to be conducted to observe the chip shape using high cutting speeds which will provide a better comparison between the chip shape at higher cutting speeds.

The aim of the present study was to compare the basic chip shape produced in experiments and numerical simulations. From Figure 9.19, it can be seen that the chip shape in both the cases was comparable with regard to the curvature. Continuous chip was observed to be produced in FE simulations as well in the experiment. In numerical simulations, the chip formation process was observed from the initial contact between the cutting tool and work piece at \( t = 0 \) to a state of the fully formed chip \( t = 0.005 \text{ s} \) (see Figure 9.19b). Figure 9.19a shows the image taken 5 second after the cutting was initiated in the experiment. Material separation in front of the cutter and the slipping of the material along the cutting plane were not observed which requires close examinations at the cutting edge of the tool and should be investigated in future.

### 9.5.2 Comparison of cutting forces

To validate the developed model of bone cutting, experimental measurements of plane cutting forces for real bone tissue were conducted. The bone used in plane cutting experiments was from the same animal and anatomical location as investigated in the material data acquisition tests for FE modelling. The simulation time for cutting was smaller than the experimental one because of contact and divergence problems. To compare, quantitatively, the cutting forces obtained from the FE model with experimental results, an additional series of numerical simulations was implemented for lower velocities (up to 16 mm/s) as used in experiments. A comparison of calculated and measured cutting forces is shown in Figure 9.20 for three values of DOC. It is evident that two sets of results
were in good agreement. In case of CC, a difference of 16.6%, 7.5% and 12.5% between the measured and predicted values was found, respectively, for DOC of 0.1 mm, 0.2 mm and 0.3 mm. In UAC, the difference between the measured and calculated values was 31.3%, 8.7% and 28% for DOC of 0.1 mm, 0.2 mm and 0.3 mm respectively.

![Comparison of calculated and measured cutting forces for various values of DOC](image)

**Figure 9.20** Comparison of calculated and measured cutting forces for various values of DOC. \( \psi = 10 \, \mu m, \gamma = 20^\circ, f = 20 \, \text{kHz}, \ a = 10 \, \mu m, V_c = 10 \, \text{mm/s} \)

### 9.5.3 Comparison of cutting temperatures

FE results were compared to temperature values obtained from the drilling experiments. Results of a 2D FE model may be compared to a cutting process such as turning of bone, where a single-point cutting tool is utilised, but the shape of bone restricts such operation. In turning, measurement of cutting temperatures at the cutting zone can be measured using a remote measurement technique such as infrared thermography, since the tool tip would be exposed to the thermal camera. In drilling operation, however, the measurement of temperatures is complicated as the drill cutting edge is hidden to the thermal camera. The FE results were compared to the results obtained from bone-imbedded thermocouple system due to the obtained values of temperatures in bone being consistent with those reported in literature. Also the temperatures were obtained at the work piece in FE simulations, which was similar to the measurements in bone-imbedded thermocouple system. To compare the calculated and measured temperature growth in the bone tissue, FE analyses were run by assigning the initial temperature of 25°C (the initial temperature of the bone as measured in experiments) to the work piece with no cooling condition applied.
The linear speed of the cutting lip of a drill at the outer diameter was calculated using Equation 8.1 and the obtained values (second column of Table 9.1) were used in simulations. Here, \( \Delta T \) represents the rise in temperature in the bone material above the initial temperature of 25°C. The temperature was measured as a function of the drilling speed and compared with the results obtained from the 2D FE cutting model. The maximum temperature rise showed an increase with the increase in cutting speed, which was consistent with experimental findings. For the range of drilling speeds used in the experiments, the variation in each data set was noted to increase with an increase in drilling speeds (see Table 9.1). The reason for those variations are unknown to the author at this stage of investigation. The difference between the experimental and calculated temperature values, for the drilling speed of 3000 rpm, was about 21%.

Discrepancies between the calculated and measured values may be attributed to assumptions in the FE model such as material homogenisation as well as the use of reported thermal properties of cortical bone tissue taken from the literature. In temperature measurements, an obvious reason for the difference was that the measurements were obtained from the thermocouple that was placed at a distance of 1 mm from the cutting edge of the drill in the experiments. The difference may also be due to the fact that the FE model relied on a constant value of friction at the tool-work piece, while in drilling experiments, it may change with cutting speed and feed rate. However, given all of the inherent complexities in predicting and measured quantities over a range cutting parameters, the developed model was reasonably accurate.

**Table 9.1** Comparison of temperature rise in bone as obtained from experimental drilling tests and FE calculations. Experimental results are for five drilling tests for each speed \((D = 4 \text{ mm}, \ h = 0)\)

<table>
<thead>
<tr>
<th>Drilling speed, ( N ) (rpm)</th>
<th>( V_c = \frac{\pi DN}{60} ) (mm/s)</th>
<th>( \Delta T ) (°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td><strong>Experimental</strong></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Mean (SD) [Range]</td>
</tr>
<tr>
<td>600</td>
<td>126</td>
<td>5 (1.87) [3-7]</td>
</tr>
<tr>
<td>1000</td>
<td>209</td>
<td>20.2 (3.1) [17-25]</td>
</tr>
<tr>
<td>1400</td>
<td>293</td>
<td>38.6 (2.79) [36-42]</td>
</tr>
<tr>
<td>1800</td>
<td>377</td>
<td>50.8 (3.35) [47-55]</td>
</tr>
<tr>
<td>2200</td>
<td>461</td>
<td>58.4 (3.05) [55-61]</td>
</tr>
<tr>
<td>2600</td>
<td>544</td>
<td>63.2 (4.20) [58-68]</td>
</tr>
<tr>
<td>3000</td>
<td>628</td>
<td>67 (4.5) [60-72]</td>
</tr>
</tbody>
</table>
9.6 Conclusions

A fully parametric 2D finite element model was developed to simulate the bone cutting process. The model can simulate the cutting process for various cutting parameters, such as DOC, cutting velocity, tool’s vibrational frequency and amplitude. The model allowed the interaction between the bone and cutting tool to be studied, hence enabling the evaluation and optimization of the cutting procedure. The chip formation process was studied to predict distributions of stresses, cutting forces in conventional cutting (CC) and ultrasonically-assisted cutting (UAC). The model was used to analyse the temperature in the bone material in CC only. The Jonson-Cook material model with the known material properties was used in the cutting simulations.

When subjected to ultrasonic vibration, the material experienced a varying stress state depending on the motion and location of the tool in each cycle. The average over the cycle stress level in the material in UAC was lower than that produced in CC. The level of cutting force in numerical simulations was affected by tool geometry and the depth at which the tool cuts the bone. The material removal rate and extent of lubrication were the main factors responsible for the variation of the cutting force. The model was an attempt to provide basic understanding of the relationship between the process parameters and tool geometry, on the one hand, and the cutting force, on the other. In simulations, the cutting speed within the studied interval was found to have no effect on the cutting force, which was consistent with our experimental results for lower cutting speeds. A non-permanent contact in the case of UAC was the reason for its force diminishment as compared to CC.

The results of FE analysis can be used to optimize bone surgery in order to achieve minimum invasion. From this study, the low depth of cut and a cutting tool with a sharp edge, which reduces the cutting forces, can be recommended for orthopaedic bone surgery. UAC can be recommended for bone cutting if a high frequency and amplitude up to 30 µm is used. The additional temperature load of the vibrated tool on bone tissue should be studied and can be reduced (if necessary) by using appropriate cooling conditions. Research into the bone cutting temperature is crucial for bone surgery, in addition to a study of the accuracy of the shape of the obtained cut (it may affect the strength between the bone and fastening components) and the level of cutting force to allow smooth penetration.

A thermo-mechanically coupled FE model of bone cutting was used to simulate the temperatures during the high-speed cutting process. Higher cutting speeds of drilling were observed to be responsible for inducing thermal necrosis in the bone tissue. A saturation of the level of maximum temperature with the speed was not observed for the range of cutting
speeds studied and can be examined in additional experimental tests. By employing conditions corresponding to the use of the coolant, it was possible to demonstrate an increase in the allowable level of cutting speeds, that did not induce temperatures over the necrosis threshold by 67%. In case of adequate cooling, cutting speeds up to 550 mm/s may be used without the risk of inflicting thermal injury to the bone tissue. This cutting speed correspond to a drilling speed of approximately 2600 rpm for a drill with 4 mm diameter. The temperature rise and necrosis penetration were noted to have been affected by the cutting speed, depth of cut and bone thermal properties. Appropriate control of cutting parameters will allow prevention of the thermal necrosis and hence could improve bone regeneration and its stronger attachment to the implant. The homogenised FE model was validated by comparing the measured and calculated values of cutting force and temperatures and were found in agreement. The differences in the calculated and measured values can be minimised by providing more accurate material data of cortical bone to the FE model.

Currently, our finite element model relies on the bone thermal properties reported in the literature; they should be obtained experimentally in future for the same bone as used in experiments. Analysis of the mechanics of cutting and its effect on stress distributions in the bone tissue and the cutting force and temperature calculations would facilitate optimal prosthesis design and fabrication. This may also be helpful in reducing the risk of mechanical failures and in improving bone-implant longevity.
Chapter 10. Conclusions and Recommendations for Future Work

10.1 Conclusions

The thesis was mainly aimed at understanding mechanics of the bone cutting process. The aim was achieved by conducting a comprehensive literature survey followed by a series of experiments and numerical simulations. Various experiments were carried out in order to quantify the mechanical properties of bovine cortical bone for two purposes: (1) mechanical tests to characterise mechanical properties of bone in order to develop a FE bone cutting model; (2) bone cutting experiments to investigate the effect of various cutting parameters on the cutting force and temperature and study the benefits of ultrasonically-assisted cutting (UAC) technique in orthopaedics. Based on results of those experiments, a fully parametric and thermomechanically coupled 2D FE bone cutting model was developed to simulate the bone cutting process. That numerical model allowed predictions of stress distributions in bone, the tool penetration force and temperature in the cutting zone. Most importantly, the developed model adequately reflected experimental results. Following the list of objectives set in Introduction (see Chapter 1), the PhD research study has brought the results described below.

10.1.1 Material characterisation

Experiments were planned as a continuation and extension to those already performed by other researchers on the in-house UAT setup. The new studies performed included the high-speed filming of the bone drilling process to visualise chip separation in CD and UAD. Tension tests were conducted to acquire the stress-strain relationship and strain rate sensitivity for developing a FE bone cutting model.

Tension tests

Tension tests were conducted to study the stress-strain relationship for fresh cortical bone specimens obtained from the mid-diaphysis of a bovine femur. All the specimens were tested in the direction parallel to the longitudinal axis of the bone. Tension tests revealed an inelastic behaviour consistent with that reported before. After yielding, the specimens showed a bi-linear hardening up to an average failure strain of 2.4%. The specimens were tested in the strain rate interval between $0.00001 \, \text{s}^{-1}$ and $1 \, \text{s}^{-1}$. The yield stress was rate-dependent which provided an opportunity to utilize a well established Johnson-Cook material model, originally suggested for metal cutting, in the numerical
model of bone cutting. Material separation was based on the failure stress obtained from the obtained stress-strain relationship.

**Nanoindentation test**

The elastic modulus is an essential material property used in all analytical and numerical modelling schemes of the material’s mechanical behaviour. The modulus was obtained from a nanoindentation test made at the osteonal and interstitial bone. The average elastic modulus of the interstitial bone was about 10% higher than that of the osteonal bone. This technique did not result in quantitative data on residual stresses in the sub-surface layer of the samples caused by cutting followed by different grades of polishing. The value of the elastic modulus measured in nanoindentation test was close to that obtained from the stress-strain relationship of the similar bone. Due to the small variation of the elastic modulus, the composite structure of the bone was idealised as homogeneous with the effective elastic modulus calculated using the rule of mixtures. The effective elastic modulus was used in numerical simulations of bone cutting.

**10.1.2 Experimental study of bone cutting**

**High-speed filming**

Dynamics of the drill-work piece interaction was studied using high-speed (HS) filming, which underpinned the in-depth analysis of both CD and UAD. Results of HS filming showed the chip formation in a detailed view with a frame rate up to 5000 frames per seconds. Those results were helpful in understanding mechanics of the drilling process by visualising the chip formation. Observations of results from high speed filming provided necessary information on the chip formation process, which affected mechanics of drilling. The formation of continuous and segmented chips was found to be dependent on the cutting mode of the drill - conventional or ultrasonic. The study will enable the development of a 3D FE bone drilling process for both CD and UAD techniques. It was noticed that UAD improved the process of chip evacuation from the cutting zone by breaking chips into small pieces. Those small/segmented chips were later found to decrease the drilling thrust force and resulted in higher surface quality of holes. It is also expected that such small chips will produce lower heat generation, reducing the risk of thermal necrosis. Interestingly, UAD of fresh bone produced chips similar to those produced by CD of dry bone.
Surface roughness measurements

The study of surface roughness was motivated by the chip formation process that resulted in CD and UAD as observed using high-speed filming. Surface roughness analysis of the drilled holes had been performed using contact and non-contact techniques. Both techniques have proved the significance of UAD in improving the surface roughness as compared to CD. The measured values of the roughness will help in the development of more advanced numerical and analytical models to investigate strength of the fastening devices anchoring the bone.

Measurement of cutting force

- The cutting force in conventional plane cutting (CPC) of bone ranged from 17 N to 83 N for various combinations of the process parameters. The force measured in conventional drilling (CD) ranged from 20 N to 65 N. The force in ultrasonically-assisted plane cutting (UAPC) and ultrasonically-assisted drilling (UAD) was from 1.5 to 3 times lower compared to that in CPC and CD.
- The parameter, most influencing the cutting force in plane cutting, was the depth of cut (DOC). In plane cutting, a lower cutting speed (up to 16 mm/s) was used due to the natural limitations in bone dimensions. The cutting force was not affected by the range of cutting speeds tested. Unlike plane cutting, the drilling thrust force and torque dropped with an increase in the drilling speed in both CD and UAD. A linear relationship between the feed rate and the thrust force was found in CD, while in UAD the effect of the feed rate was negligible. This fact led to the conclusion that, while using UAD, a higher feed rate should be used to reduce the osteotomy/surgery time if an efficient irrigation system is applied to the drilling site to minimise the risk of thermal damage.
- The force in plane cutting and drilling was significantly reduced by changing the frequency within the range used. That, however, did not influence the force for DOC above 0.2 mm in plane cutting. From that study it was concluded that the application of ultrasonic frequency was more beneficial when lower DOCs were used in plane cutting. In drilling, the torque was also measured to drop with the increase in frequency. The imposed vibration on the tool will allow the surgeon to apply lower force levels, which will minimise the risk of unnecessary damage to the bone tissue as well as the breakage of the tool itself.
- The change in the ultrasonic amplitude between values ranging from 5 µm to 15 µm reduced the force in plane cutting as well as in drilling. There was no benefit (force reduction) of increasing the amplitude above 15 µm. A similar trend was found in
torque measurements. The study suggests an amplitude up to 15 µm with the maximum frequency tested (30 kHz) for minimal invasive cutting of bone.

**Measurement of cutting temperatures**

- Temperatures were not measured in UAD due to its effect on the response of a thermocouple in the presence of imposed vibration on the tool. The obtained readings were observed to fluctuate between unrealistic values. This can be covered by the use of a remote sensing technique such as infrared thermography and should be the focus of future studies. Two methods were applied to assess the temperature during the drilling process i.e. a drill-imbedded thermocouple system for measurement at the drill site or a bone-imbedded thermocouple system for measurements in bone. The results obtained from the study were consistent with those reported in literature. Those measurements were also verified by our FE model.

- The maximum temperature change measured by a thermocouple placed in the bone ranged from mean values of 6°C to 90°C for drilling through a single cortical thickness of approx. 10 mm and up to a depth of 40 mm using various combinations of the process parameters. For the same drilling parameters (except a difference of 1 mm in the diameter of the drill used in two the techniques), the tool-imbedded thermocouple system measured considerably higher temperatures ranging from 65°C to 130°C. As the temperature values obtained from the bone-imbedded thermocouple system were more close to those reported in literature for nearly similar drilling parameters and conditions, the study recommends this method for temperature measurements in bone.

- The parameters most significantly affecting the temperature increase in the bone, were the drilling speed and feed rate. The study concluded that the drilling speed should be kept as low as possible together with a higher feed rate in the presence of efficient cooling to minimise the osteotomy time. According to the measurements with the bone-imbedded thermocouple system, the minimum drilling speed, to avoid thermal damage, should be kept below 1600 rpm for a drill diameter of 4 mm and the feed rate 50 mm/min.

**10.1.3 Numerical simulations of bone cutting**

Two-dimensional thermomechanically coupled finite element models of both CC and UAC have been developed using the FE code MSC.MARC. It was the first attempt to
study the cutting process of bone using FEM. The developed FE models present several features and limitations described below.

- The models were fully parametric based on the mechanical properties obtained in the tests and thermal properties taken from the available literature.
- The models were thermomechanically coupled, capable to simulate the tool penetration force and temperature at the cutting zone.
- The models predicted the stress and temperature distributions close to the cutting edge which was not possible with the experimental method.
- The chip separation was not based on the pre-defined line or use of a notch, and the material separation mechanism was driven by more physical criteria based on the magnitude of the failure stress obtained in the tension tests. It made it possible to use the same model to analyse different tools, which, otherwise, would have required defining a specific notch for every new tool.
- A frictional condition was modelled between the tool and chip.
- Automatic remeshing and rezoning were involved.
- The amount of heat transfer from the work piece to the tool is dependent on the contact heat transfer coefficient which is a complex parameter to be measured experimentally. This restricted the author to model heat transfer into the tool which would require a reliable value of the heat transfer coefficient. In UAC, the intermittent contact between the work piece and tool make the transport of heat across the tool-work interface a very challenging task. In this study, the temperature measurements were not simulated for UAC and should be covered in future studies.

**Thermomechanics of bone cutting**

The following section presents the conclusions from the FE simulations.

- The distribution of stresses in bone during a cutting process was simulated, which otherwise would be difficult to measure experimentally. Those calculations may be useful in understanding the functional relationship of bone with fastening devices. This will promote the development of numerical models for further simulations of osseointegration.
- The model predicted a considerable drop in the cutting force, when a sharper cutting edge was modelled. A tool with a smaller nose radius is recommended to be used for bone cutting. The larger DOC resulted in higher cutting forces due to the increased material removal. Cutting forces also decreased for lubricated cutting conditions as compared to dry cutting.
The intermittent character of the chip – cutting tool interaction determined the main differences in the stress distribution for CC and UAC. FE simulations demonstrated a transient stress state in UAC during a single cycle of the tool vibrations and nearly quasistatic one for CC. The type of stress distribution in the penetration stage of a single cycle of vibration in UAC was somewhat analogous to that of CC, However, stress levels in UAC were considerably lower than those in CC for about 57% of the cycle time. As a result, the averaged values of stresses obtained in simulations of UAC were significantly lower than those in CC.

For the same cutting parameters (cutting speed, depth of cut and friction) and a typical type of ultrasonic vibration used, the calculated cutting force in UAC was 10-65% of that in CC, which was comparable to the results of plane cutting experiments.

The model successfully predicted the temperature close to the cutting edge for several cutting parameters. Higher levels of the cutting speed and DOC were the main parameters for inducing necrosis. The study suggests a cutting speed lower than 300 mm/s (without cooling) and 550 mm/s (with cooling), which correspond to drilling speeds of 1400 rpm and 2400 rpm, respectively, for a drill of 4 mm diameter.

10.2 Future Work

The need for further work follows from the results of this thesis.

Experimental work:

The stress-strain relationship of the bone was obtained using a specimen excised from the mid-diaphysis and the direction of loading was along the longitudinal axis of the bone. This was the only possible direction in which the specimen could be taken to provide a sufficient length for gripping between the holders of the tension test machine. More experiments are needed to test specimens along the radial and tangential directions. This will allow the provision of additional data to develop a FE bone cutting model with anisotropic material properties.

It would be of interest to carry out the high-speed filming experiments with higher magnification to see smaller details of the chip formation mechanisms in drilling. The equipment used in high-speed filming can provide a maximum of 5000 frames per second. Chip formation mechanism in UAD may be further studied using a higher filming rate ($10^5$ frames/s or higher), as this would allow observation of the
cycle of ultrasonic vibration. This will help in developing a FE model for segmented chip formation under superimposed vibrations on the tool.

- Experiments can be conducted to measure the surface roughness using drilling parameters other than those used in this work as well as to study their effect on the surface quality. It will also be useful to measure the surface roughness of the holes drilled along the direction parallel to the bone longitudinal axis.
- Experiments can be performed to measure temperature in UAD using a remote sensing technique to avoid the errors encountered with thermocouple readings in the presence of ultrasonic vibrations. It will be however, difficult to sense the temperature at the cutting zone when the drill bit penetrates into the bone and its cutting edge is not visible to the infrared device. A possible approach is to measure the instantaneous temperature the moment the drill exits the cortical tissue.
- All the tests pertinent to bone cutting were carried out without cooling. Experimental work is required to investigate the influence of irrigation with various flow rates on the temperature rise in bone, which will reproduce the conditions similar to that used in real orthopaedic surgery. Such in-vitro studies would provide valuable data on cortical temperatures incurred similar to clinical conditions and will serve as a means of comparison with the calculated temperatures obtained from our FE model.
- In this study, drilling temperatures were measured for cortical bone only. Experimental studies are needed for such measurements in trabecular bone.
- A complete set of experiments can be planned using the suggested set of optimum parameters from simulations performed in this thesis. This can provide an additional comparison for the obtained results of simulations from this thesis.

**Numerical simulations:**

- The current model was based on homogenisation of the bone composite microstructure, with an effective elastic modulus, averaged from the elastic moduli of the bone constituents (osteonal bone and interstitial bone) used. The model can be improved by incorporating anisotropy, which will predict the cutting force along the desired anatomical directions.
- The geometry of the FE model was limited to a height of 1 mm and width 2 mm to avoid the use of a large memory space. This restricted the measurements of temperature at a location 1 mm from the cutting edge of the cutting tool. To
compare calculated temperature with experimental results a larger model that is geometrically consistent with the real drill-bone is needed.

• The thermal properties the bone in the model were not obtained experimentally and rely on those from the literature. A more realistic model can be developed provided the mechanical and thermal data necessary to simulate the thermomechanics of the cutting process are obtained for the same bone.

• Based on experimental evidence, a FE model is to be developed simulating the cutting temperature in UAC. This may be possible by introducing the contact heat transfer at the tool-work interface for intermittent contact condition.

• There is a need to produce a 3D model of the drilling process to avoid the assumption of cutting process of drilling as a 2D one at the drill corner level. The calculated temperatures will be compared with those obtained from our drilling tests.

• FE modelling with a finer mesh over the whole work piece can allow the study of the surface roughness.

• An improved re-meshing algorithm in FE software can allow more consistent predictions of forces and the application of boundary conditions directly on the deformable bodies as well.
References


