

Ultrasonic tool to cut human bone: Cutting speed and forces necessary for potential remote robotic arms

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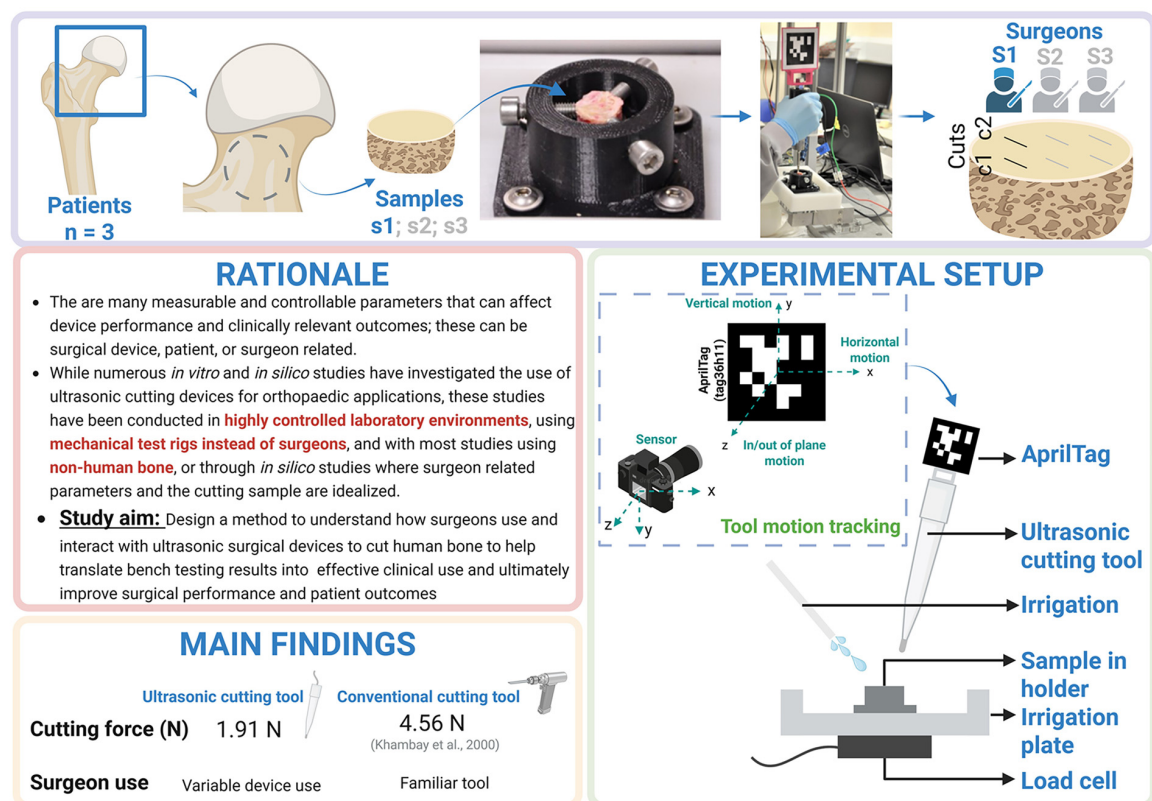
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Graphical abstract


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Abstract

Procedural-related musculoskeletal pain is common among orthopaedic surgeons, often caused by the repetitive use of high-force bone-cutting tools. Ultrasonic cutting devices, which can operate with lower force, may help reduce this physical burden. In this study, three practising orthopaedic surgeons each performed two cuts on three fresh cortical bone samples, harvested from excised femoral necks from three patients undergoing hip replacement surgery. The study was conducted using an ultrasonic cutting device in a controlled yet clinically reflective environment. A novel setup captured real-time data on surgeon-related parameters, including vertical cutting force and vertical and horizontal cutting speed. Consistent with previous research, we confirmed that ultrasonic devices enable low force cutting (average 1.91 N). However, our findings revealed significant variability in how each surgeon interacted with the device – including how much force each surgeon applied, and how the device was manoeuvred which can influence device performance, thermal effects, and overall clinical outcomes. Given the critical importance of surgeon-related factors, our results highlight the need to understand how each surgeon interacts with these devices differently. This insight can inform training and device optimisation strategies; help translate bench testing results into effective clinical use and ultimately improve surgical performance and patient outcomes. Additionally, our findings support the potential benefits of integrating ultrasonic devices with robotic platforms to maintain consistent cutting parameters. Future research should investigate optimal cutting parameters, evaluate different blade profiles, assess result generalisability and compare outcomes before and after training or system enhancements.

Keywords

orthopaedic procedures, biomedical devices, robotic assisted surgery, biomedical instrumentation, implants/prosthetics

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Introduction

Orthopaedic surgeons face many occupational hazards during their practice, including procedural-related musculoskeletal pain and hand arm vibrating syndrome,¹ which is a known risk factor for developing carpal tunnel syndrome.² A recent survey by McQuivey et al. demonstrated high prevalence of procedural-related musculoskeletal pain among arthroplasty surgeons,³ which may occur due to repetitive use of bone saws during surgery. Ultrasonic bone surgery devices were first commercialised for oral and maxillofacial surgical procedures to perform osteotomies,^{4–21} then were later extended to neurosurgical^{22–26} and orthopaedic^{27–30} surgeries. These devices cut bone tissue with low force, which helps to reduce the physical burden on surgeons during surgery and also reduces their vibration exposure as ultrasonic vibrational amplitudes are insignificant compared to those produced by a mechanical saw.³¹ Compared to other surgical instruments, ultrasonic devices boast many advantages that have the potential to accelerate post-operative bone healing, including reduced cutting force,³² increased precision,^{26,33} controlled blood loss^{23,24} and reduced collateral damage to surrounding tissues.^{23,26} Furthermore, while surgical robotic technology has not yet reached a level where completely replacing surgeons is feasible,^{34,35} many common orthopaedic procedures such as total hip arthroplasty (THA) and total knee arthroplasty (TKA) are increasingly being performed with robotic assistance.^{36,37}

However, as ultrasonic surgical devices largely rely on making multiple small, shallow cuts,³⁸ which enables precise cuts in small bones, this limits their applications

to orthopaedic large bone surgeries where large cuts are required. There are many measurable and controllable parameters that can affect device performance and clinically relevant outcomes, which can be largely categorised as surgical device related, patient related or surgeon related. Device related parameters include ultrasonic transducer design and blade design (including blade material, vibration frequency and amplitude, and cutting-edge geometry). Patient related parameters include bone density, bone type, bone orientation and state of the bone, including hydration or moisture content and temperature. Finally, surgeon related parameters include cutting force, cutting speed, length of cut, angle of approach, the time that the device is in contact with the bone sample being cut and number of passes of the surgical device on the bone sample.

While numerous *in vitro* studies have investigated the effect of ultrasonic cutting on bone stress (to minimise bone mechanical damage),^{39–45} reduce cutting temperature^{46–52} and improve device performance, including blade design,^{53,54} device design,⁵⁵ transducer design,^{56–59} and cut parameters,^{60,61} these studies have been conducted in highly controlled laboratory environments using mechanical test rigs, and with most studies using non-human bone (bovine,^{46,48,50,62,63} porcine,^{43,45,49,61} synthetic⁶⁴ or specific bone not mentioned⁶⁰). Alternatively, *in silico* modelling has been used where input cutting parameters (normally surgeon related parameters) and the cutting sample are idealised.^{38–42,47,51} While highly controlled lab-based experiments are critical to understand how to optimise surgical device operation, they do not reveal how different surgeons interact with these devices under the

complex and often unpredictable conditions encountered in clinical practice. Understanding how surgeons use and interact with ultrasonic surgical devices can provide insights that help translate bench testing results into clinical use.

In this study, we present a novel experimental setup, and demonstrate feasibility of the setup, that allows practicing orthopaedic surgeons to perform cuts on fresh human cortical bone using an ultrasonic surgical device within a controlled yet clinically realistic environment. The setup can capture surgeon related parameters, including applied cutting force, time-stamped tool trajectory from which the horizontal and vertical components of cutting speed can be calculated, and bone penetration rate. Our results represent an

approach that facilitates detailed analysis of surgeon–device interaction to inform future strategies for training and optimising device usage, using information gained from controlled lab tests. The ultimate aim is to improve surgical performance and patient clinical outcomes.

Methodology

We set up a robot operating system (ROS) coding suite⁶⁵ to record and synchronise time stamped surgical device position and vertical force from the Arduino microcontroller.

To record the cutting force, a custom designed test clamp was connected to an irrigation plate, which was mounted on a 6-wire single point load cell (1000 N capacity) and the assembly was fixed to a workbench. To measure force, the load cell was connected to an Arduino Due microcontroller for processing. To track the surgical device position, an AprilTag (tag36h11) was mounted onto the ultrasonic surgical device using a custom 3D printed tag holder^{66–68} (Figure 1).

To visualise the tag, a SpotLight Pro Webcam with LED lights (Trust International B.V.) was used in conjunction with the `usb_cam` (a ROS driver for V4L USB Cameras),⁶⁹ which had been calibrated using the Kalibr Calibration toolbox^{70,71} using a Aprilgrid with 36 AprilTags (6 × 6) printed and glued onto a rigid piece of cardboard. For calibration, a pinhole-radtan model was chosen and the camera frame rate was set to 4 Hz. The maximum reprojection pixel error was 1.2 with most of the assessed data points lying below 1 pixel (Figure 2).

Ultrasonic bone cutting devices operate at a low ultrasonic frequency (typically between 20 and 35 kHz).⁵⁷ In a preliminary study, we compared a 20 kHz and a 35 kHz device and found that the 35 kHz device exhibited less damage to the bone tissue at the cut site.⁷² Therefore, in this study we used a 35 kHz device with a previously described rounded blade.⁷³ Detailed description of the surgical device design and characterisation, including measurement of the ultrasonic vibration response, have been reported previously.³⁰ Briefly, three orthogonal components of vibration velocity were measured at a grid of points on the device surface using a 3-D laser Doppler vibrometer.

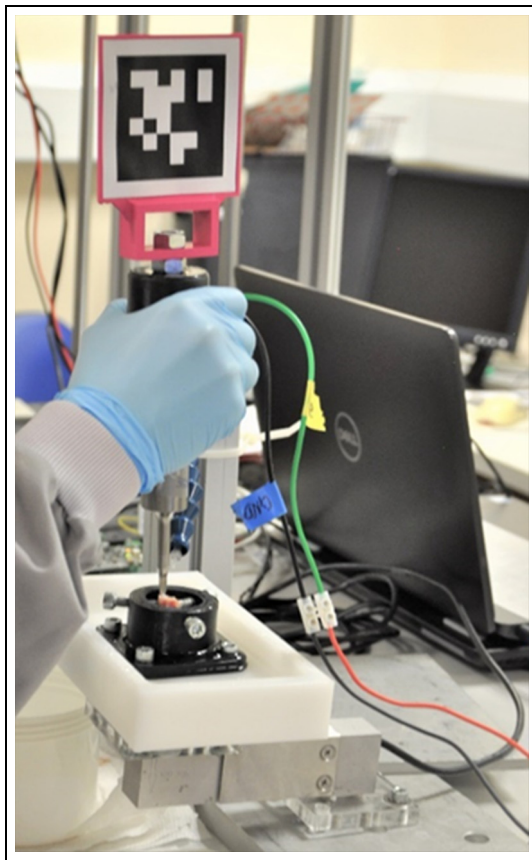


Figure 1. Experimental setup showing orthopaedic surgeon cutting a human cortical bone sample using a custom ultrasonic tool, tracked using an AprilTag.

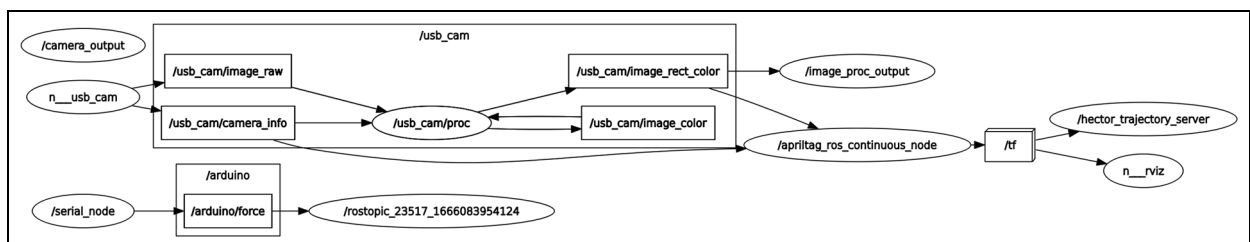


Figure 2. A plot demonstrating the ROS system (nodes and topics) used in this study.

Frequency Response Functions (FRFs) were computed from excitation and response signals using dedicated software, followed by curve fitting and modal analysis to extract FRF magnitude and phase and hence the modal parameters.³⁰ During cutting, the device's electro-mechanical characteristics of impedance, resonance frequency and consumed acoustic power were measured using a commercial ultrasonic generator and resonance tracking unit (PDUS210 Piezodrive).

The ultrasonic device was driven by the generator with 800 Vp-p output voltage and power load of 210 W, equivalent to a surgical device tip displacement amplitude of 36 μm . While this is lower than some of those reported for commercial surgical devices, ranging from 40 to 60 μm ,⁷⁴ it has been determined in experiments that cortical bone can be cut effectively with a surgical tip displacement amplitude of 20 μm .

Force calculations

The signal from the force transducer was connected to an Arduino microcontroller via a strain gauge amplifier, which provided both signal amplification and analogue-to-digital (A/D) conversion. The force measurement range, denoted as MR_F , was determined using the sensitivities of the force transducer (SR_T) and amplifier (SR_A), which define the sensitivity ratio (R), and the full-scale transducer range (MR_T) as shown in equations (1) and (2).

$$R = SR_T / SR_A \quad (1)$$

$$MR_F = MR_T / R \quad (2)$$

SR_T is the force transducer sensitivity (2 mV/V)

SR_A is the amplifier sensitivity (0.11 mV/V)

R is the sensitivity ratio

MR_T is the full-scale force range of the transducer

MR_F is the calibrated force measurement range (55 N)

The strain gauge amplifier output was configured to a range of 0–10 V, with the Arduino Due only capable of reading voltages from 0 to 3.3 V. The amplifier gain was adjusted accordingly so that a force of 55 N corresponded to 3.3 V at the Arduino input. To convert voltage readings into force, the resolution of the load cell at 0 and 1 N was first calculated (equation (3)), followed by the actual force calculation using the ADC readings (equation (4)):

$$\text{Res}_{\text{LC}} = \frac{1}{(F_{D,1} - F_{D,0}) \text{Res}_{\text{ADC}}} \quad (3)$$

$$F = [F_D - F_{D,0}] \text{Res}_{\text{ADC}} \text{Res}_{\text{LC}} \quad (4)$$

Res_{LC} is the load cell resolution calculated for 1 N force in (N/V).

Res_{ADC} is the resolution of the analogue to digital converter, in this case $3.3 \text{ V} / 2^{12} \text{ bits}$.

F is the converted force reading in N.

F_D is the digital force reading in bits.

$F_{D,0}$ is the digital force reading when there is no load on the load cell in bits.

$F_{D,1}$ is the digital force reading when there is 1 N load on the load cell in bits.

Eliminating the force of the irrigation fluid

As irrigation was used during the experiment, it was required to eliminate the force of the irrigation fluid from the force readings. The force during 1 min (equivalent to each bone cutting experiment duration) of irrigation with PBS was measured and the average force and standard deviation was calculated (equation (5)).

$$F_1 - F_2 = (\mu_{F_1} - \mu_{F_2}) \pm \sqrt{\sigma_{F_1}^2 - \sigma_{F_2}^2} \quad (5)$$

F_{irr} is the average force reading for 1 min of irrigation fluid weight (N)

F_1 is the average force reading before accounting for F_{irr} (N).

$F_2 = F_1 - F_{\text{irr}}$ is the average force reading after accounting for F_{irr} (N).

μ is the calculated mean for the data set (N).

σ is the calculated standard deviation (N).

The measured average force due to irrigation, for a 1 min duration, was $0.38 \pm 0.14 \text{ N}$.

The measured average force for each experiment is also calculated using equation (5), however, F_1 (N) is calculated over the total recorded time for that experiment.

Measuring cutting parameters during device operation by a clinician.

Cutting tests

During cutting tests, samples were irrigated with Phosphate-Buffered Saline (PBS) at 25°C with the irrigation controlled using a MINIPULS® 3 Peristaltic pump set to 38 revolutions per minute. Three practising orthopaedic surgeons from the Edinburgh Royal Infirmary hospital volunteered to take part in the study. Each clinician was instructed to find a comfortable position at the workbench, which allowed an adequate arm rest without contacting the assembly. All surgeons chose to perform the cuts in a standing position, reflecting as closely as possible their clinical handling of a cutting instrument in surgery. Before performing the cuts, each surgeon was given a bone sample and asked to practice a few cuts to familiarise themselves with the surgical device. Once they were comfortable with using the surgical device, cortical bone samples were placed in the test clamp. Each surgeon was instructed to perform two separate cuts on three separate cortical bone samples (6 cuts per bone sample) and to ensure that each

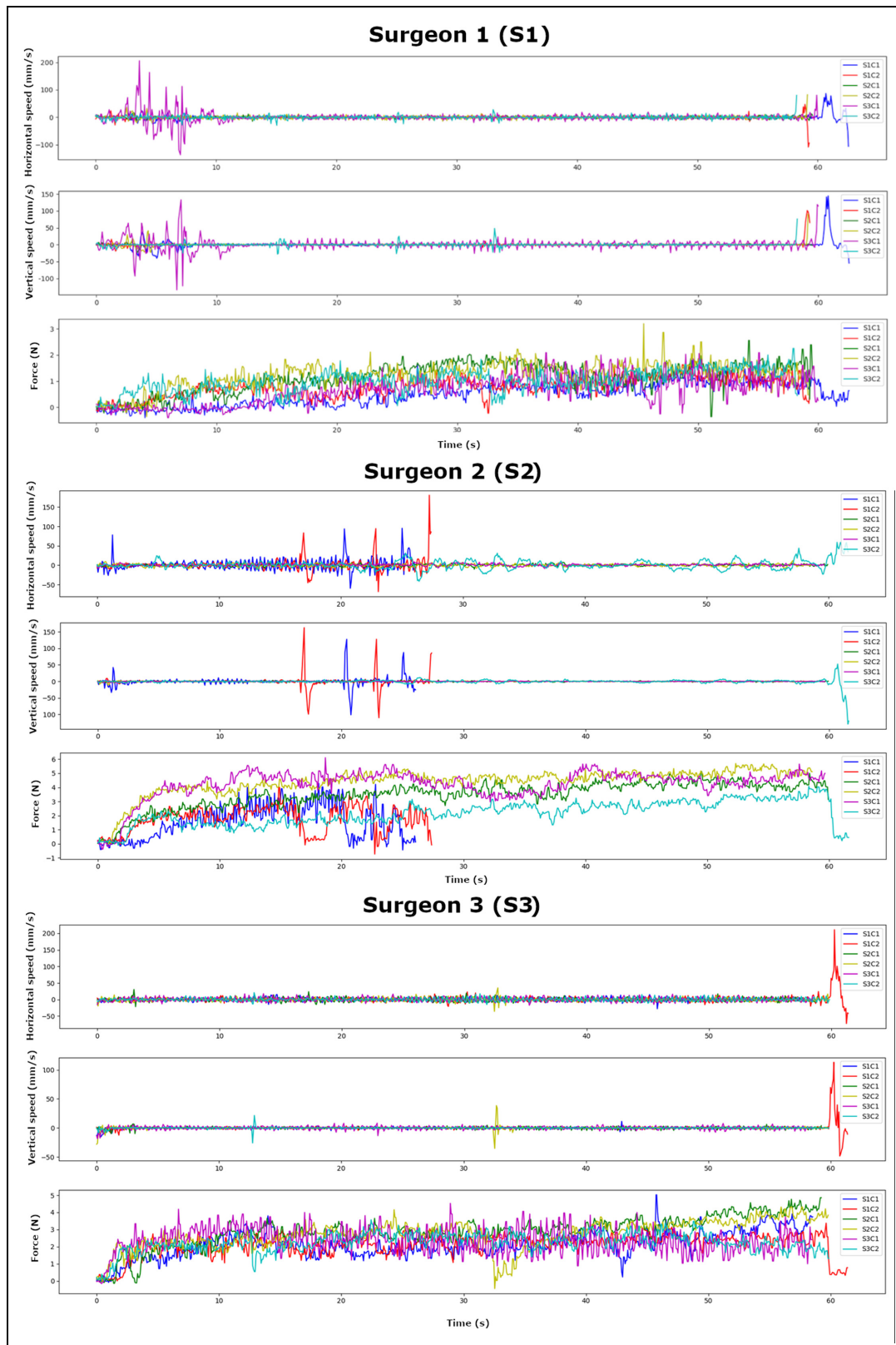


Figure 3. Horizontal and vertical cutting speeds (mm/s), and cutting force (N): Surgeon 1, Surgeon 2 and Surgeon 3.

cut was made without pausing (3 surgeons $\times$ 2 cuts $\times$ 3 bone samples, total of 18 cuts). For these tests, the following nomenclature is adopted: surgeon (S) number

(1–3), followed by sample (s) number (1–3) and cut (c) number (1–2), such that for example, S1.s2.c1 pertains to surgeon 1, sample 2, cut 1.

Cutting speed

To calculate the instantaneous horizontal (x -axis) and vertical (y -axis) cutting speed, data for the first 10 s and the last 5 s of each experiment were truncated as these were usually when surgeons moved the surgical device towards or away from the sample. Based on captured video imaging and time synchronised motion data (Figure 3), 50 mm/s was chosen as the upper limit of speed data relevant to the ultrasonic cutting of bone as it was clearly a speed above which the surgeon was moving the surgical device away from the bone, and it was not in contact with the bone samples. Data were included if load was detected (i.e. $F_2 \geq 0$ N, equation (3)) and as an extra measure, if the instantaneous speed did not exceed 50 mm/s (to ensure that readings where the surgeons stopped cutting mid experiment were omitted).

Bone penetration rate (BPR)

Unlike the device vertical speed which was measured from the AprilTag, the bone penetration rate (BPR) was calculated from equation (4):

$$\text{BPR} = \frac{C_D}{T_f} \quad (6)$$

Where:

C_D is the cut depth (mm)

T_f is the final cutting time (s)

Table 1. Bone mineral density (BMD) for the three core cortical bone samples used for cutting.

Bone sample	BMD (g/cm^3)
s1	1.24
s2	1.32
s3	1.29

For any sample with a total cutting time of less than 1 min, the BPR was not computed.

Bone mineral density (BMD) and cut depth

To remove the considerable effects on cutting tests of low bone density, we confirmed that none of the patients had any reported history of osteoporosis by measuring the BMD of the bone cores. To convert the measured Hounsfield Units (HU) into BMD, we calibrated the micro-CT scanner using two 4 mm core diameter rods of calcium hydroxyapatite phantoms with known mineral densities ($0.25 \text{ g}/\text{cm}^3$ and $0.75 \text{ g}/\text{cm}^3$) embedded in agar (1% agarose, Sigma A9539, USA) dissolved in solution using distilled water. Phantoms were imaged using the micro-CT scanner (SkyScan 1172, SkyScan N.V., Belgium) with the following settings: source voltage 80 kV, source current $120 \mu\text{A}$, exposure time approximately 885 ms, and camera pixel size of $11.57 \mu\text{m}$. Images of the phantoms were reconstructed using SkyScan software (V1.7.4.2) to generate a standard calibration curve and to enable grayscale values to be converted to volumetric BMD (mgHA/cm^2). The BMDs of the bone samples can be found in (Table 1); all were above the osteoporotic cut off volumetric BMD of $0.275 \text{ g}/\text{cm}^3$,⁷⁵ and were confirmed not to be osteoporotic.

To measure the cut depths, the bone samples were embedded in agar and imaged using the micro-CT scanner and settings as described for scanning the bone phantoms. Images of the bone samples were reconstructed using SkyScan software (V1.7.4.2). The depths of the cut were measured using stacked micro-CT scan images analysed with ImageJ software (version 1.54f). To ensure measurement consistency, the software was calibrated with the pixel-to-micron ratio of the micro-CT specifications, and the images were stacked

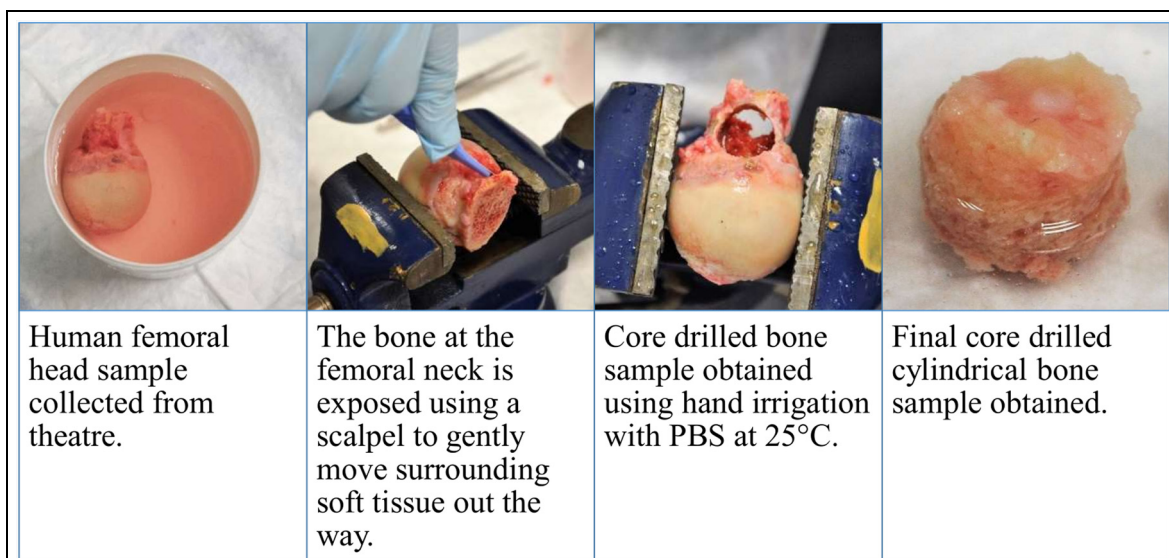


Figure 4. Process to obtain cylindrical human bone core samples.

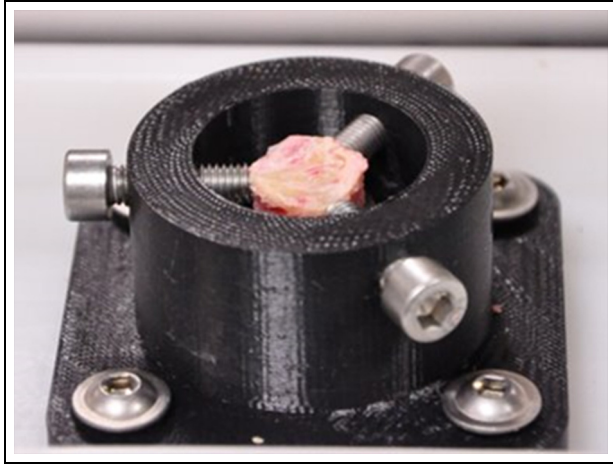


Figure 5. Human bone sample used for cutting testing held in custom designed 3D printed clamp.

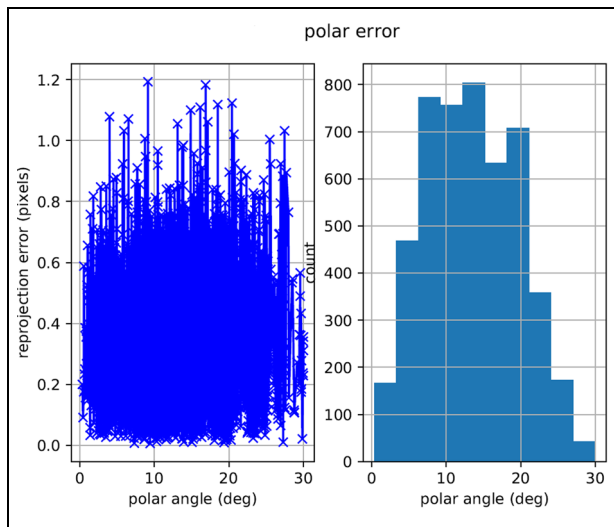


Figure 6. Re-projection error results from Kalibr calibration.

sequentially for proper alignment. Consistent regions of interest (cut site) were defined across all images using ImageJ's region of interest (ROI) manager. The depth was then calculated by measuring the distance between the surface and the bottom of the deepest point in the sagittal plane of the cut.

Testing using human cortical bone samples

Discard tissue (excised femoral heads) from three hip osteoarthritic patients undergoing hip replacement surgery was collected from the Edinburgh Royal Infirmary hospital, following approval from an institutional review board (ethics approval number: SR1342) and prior informed consent from all patients. Excised femoral heads were submerged in Dulbecco's Modified Eagle Medium (DMEM; 41965-039; Invitrogen, Paisley, UK) and stored in a sterilised container in a refrigerator (-4°C), ready for same day collection.

Following same-day sample pick-up, samples were transported to a controlled laboratory setting for experimental testing. We used a hand powered 10 mm core drill to obtain a bone sample core from the femoral neck region from each of three different hip replacement patients, where cortical and trabecular bone were deemed available (Figure 4). To ensure bone tissue integrity could be retained as much as possible, all bone samples were tested within 1 week of harvesting. Each bone sample was wrapped individually in sterile gauze soaked in Phosphate-Buffered Saline (PBS) and fresh-frozen at -20°C until required. On the day of testing, the samples were thawed for 2 h before use. For testing, cortical bone samples were mounted one at a time onto a custom designed test clamp, 3D printed in Polylactic Acid (PLA; Figure 5).

Results

The overall designed system is depicted in Figure 6.

The average horizontal cutting speed, vertical cutting speed, vertical cutting force, cutting depth, cut length and the calculated BPR across all surgeons and all samples were 4.27 mm/s, 1.64 mm/s, 1.91 N, 0.73 mm, 2.95 mm and 0.02 mm/s (Table 2).

S1 and S3 had a consistent approach to cutting bone for all samples and cuts (Figure 3). However, for S2.s3.c2 the surgeon struggled to advance the cutting blade into the bone. The surgeon pivoted the surgical device at the cut site to try to perform the cuts, which can be seen in the larger variation in the surgical device horizontal speed for this sample (Figure 3).

For S2.s1.c1 and S2.s1.c2, the resonance tracking system (Piezodrive) overloaded, which results from the surgical device being advanced too quickly into the bone, so this experiment was prematurely aborted (the cutting time was < 1 min, the force of the irrigation fluid was adjusted accordingly). For the remaining samples and cuts for S2, cuts were carried out for the full cutting time of 1 min.

Discussion

There are multiple potential advantages of using ultrasonic surgical devices for orthopaedic surgeries. In addition to the very low cutting forces and minimal vibration displacements compared to conventional bone cutting tools,³¹ ultrasonic surgical devices offer enhanced safety due to their selective cutting of mineralised bone while sparing adjacent soft tissues and the elimination of rotary cutting motion which reduces the risk of soft tissue entanglement.^{25,76} These properties also reduce the risk of bone microcracks, improve surgical visibility by limiting bleeding and debris and enable greater control in confined or anatomically complex regions. Furthermore, the applications of ultrasonic cutting tools for orthopaedic surgeries have expanded to explore the use of ultrasonic bone cutting

Table 2. Surgeon (S1, S2 and S3) vertical cutting speed (mm/s), cutting force (N) and cutting depth (mm) for sample 1 (s1), sample 2 (s2) and sample 3 (s3).

Sample no. (s)	Cut no. (c)	Horizontal cutting speed (mm/s)			Vertical cutting speed (mm/s)			Cutting force (N)			Cutting depth (mm)			Cutting length (mm)			Bone penetration rate (BPR) (mm/s)			
		Surgeon 1 (S1)	Surgeon 2 (S2)	Surgeon 3 (S3)	AVG	Surgeon 1 (S1)	Surgeon 2 (S2)	Surgeon 3 (S3)	AVG	Surgeon 1 (S1)	Surgeon 2 (S2)	Surgeon 3 (S3)	AVG	Surgeon 1 (S1)	Surgeon 2 (S2)	Surgeon 3 (S3)	AVG	Surgeon 1 (S1)	Surgeon 2 (S2)	Surgeon 3 (S3)
s1	c1	2.24	10.49	3.65	5.46	0.72	4.08	0.97	1.92	1.39	0.46	0.53	0.63	0.54	1.80	2.46	3.00	0.01	0.01	0.01
	c2	2.41	7.85	4.00	4.75	0.86	4.60	0.82	2.10	1.89	0.30	0.90	0.96	0.72	1.38	3.29	2.90	0.01	0.02	0.02
s2	c1	2.32	9.17	3.83	5.11	0.79	4.34	0.90	2.01	1.87	0.40	0.71	0.79	0.63	1.59	2.88	2.95	0.01	0.02	0.02
	c2	2.42	1.91	4.47	2.93	0.66	0.70	1.14	0.83	2.70	0.92	0.88	1.10	0.90	3.66	2.79	3.69	0.02	0.02	0.02
s3	c1	2.49	2.32	3.41	3.07	0.64	0.48	0.84	0.65	2.42	0.56	0.70	0.55	0.60	2.91	2.52	2.55	0.02	0.02	0.01
	c2	2.95	2.11	3.94	2.95	0.65	0.59	0.99	0.74	2.56	0.86	0.64	0.79	0.83	3.29	2.66	3.12	0.01	0.02	0.02
s4	c1	4.88	2.07	5.30	4.08	4.95	0.51	2.08	2.51	1.99	0.51	0.59	0.75	0.81	3.75	3.07	3.34	0.01	0.02	0.02
	c2	3.73	7.85	4.30	5.29	2.27	2.30	0.84	1.80	2.03	0.70	0.75	1.29	0.92	3.89	2.69	3.33	0.02	0.02	0.03
AVG ACROSS ALL SAMPLES		4.30	4.96	4.80	4.64	3.61	1.41	1.46	2.15	2.01	0.60	0.75	1.05	0.92	3.82	2.88	3.34	0.01	0.02	0.02
		3.19	5.41	4.19		1.68	2.11	1.11		1.91	0.65	0.75	0.89		2.90	2.80	3.14	0.01	0.02	0.02
AVG ACROSS ALL SURGEONS		4.27			1.64			1.91			0.73			2.95			0.02			

devices as a potential strategy for chondroprotection during cartilage injury repair.³⁰

However, despite the numerous in vitro and in silico studies investigating the use of ultrasonic cutting devices for orthopaedic bone cutting applications,^{39–61} these studies have been conducted in highly controlled laboratory environments, using mechanical test rigs instead of surgeons, and with most studies using non-human bone,^{43,45,46,48–50,61–64} or through in silico studies where input cutting parameters (normally surgeon related parameters) and the cutting sample are idealised.^{38–42,47,51} However, different surgeons will each interact with these devices differently, and the conditions encountered in clinical practice are complex, variable, and often unpredictable. While numerous clinical studies have investigated the use of ultrasonic cutting surgical devices for cutting various bones in the axial skeleton, including dental,^{4,5} craniomaxillofacial,^{6–13} mandibular^{11,14–21} and spinal^{22–26} bones, and in the appendicular skeleton, including bones in the hand,²⁹ these studies have typically measured clinical outcomes for example, amount of bleeding,^{4,9,14,16,19,21,23,24} precision for example, cutting length/depth,^{5,18,26,29} amount of soft tissue damage,^{6,7,15,23,26} total operative time,^{6,9,14,16,19,21,23,24,29} bone cutting time,⁷ duration of inpatient stay,^{9,16,24} post-operative healing/functional capacity^{8,16,19–21,23,29} (e.g. pain⁷⁷), blood pressure,⁹ blood values,⁹ required medication,⁹ among other clinical parameters specific to the surgical procedure,^{9–12,17} without reporting how the devices were operated. Only one clinical study tracked the tool position during surgeon cutting of facial bones in patients with a variety of pathologies. The study confirmed that tracking ensured continuous position control, however, the authors did not report the surgeon related parameters which drive overall device performance, and clinical outcomes.¹³ Understanding these interactions is crucial, as surgeon-related factors directly influence device performance and clinical outcomes. Analysing surgeon–device interaction can inform training and device optimisation strategies, helping translate bench testing results into effective clinical use and ultimately improving surgical performance and patient outcomes.

In this study, we designed a novel experimental setup that allows measuring surgeon related cutting parameters of practicing orthopaedic surgeons performing cuts on fresh human cortical bone using an ultrasonic surgical device within a controlled yet clinically realistic environment. To demonstrate feasibility of our developed setup, three practising orthopaedic surgeons each performed two separate cuts on three separate human cortical bone samples (six cuts per bone sample) (3 surgeons $\times$ 2 cuts $\times$ 3 bone samples, total of 18 cuts). As the AprilTag was located at the top end of the cutting device near to where surgeons were holding the device, our setup enabled surgeon related parameters to be captured. This analysis captured how surgeons manipulated the cutting device in real time, using the AprilTag's position and motion to

quantify the device's trajectory relative to the bone sample. Therefore, the results reflect how a practicing orthopaedic surgeon uses an ultrasonic cutting device to cut human bone samples within a controlled yet clinically realistic environment. Across all surgeons and samples, we observed an average horizontal cutting speed of 4.27 mm/s, a vertical cutting speed of 1.64 mm/s, and a vertical cutting force of 1.91 N (Table 2). These kinematic and force measurements were directly linked to micro-CT-derived outcomes, which showed an average cut depth of 0.73 mm and a cut length of 2.95 mm. From this, we calculated a Bone Penetration Rate (BPR) of 0.02 mm/s. The discrepancy is not a shortcoming of our method, but rather a revealing insight into the complex biomechanical interactions during manual bone cutting. Several contributing factors may explain the difference: (1) bone tissue exhibits both elastic and viscoelastic properties, meaning some portion of vertical tool displacement is absorbed through deformation rather than resulting in direct material removal; (2) hand and skin compliance in the surgeon's grip may introduce damping effects not accounted for in rigid-body assumptions; and (3) the cantilever-style load cell used for force measurement may introduce minor mechanical compliance. However, perhaps most important, our tracking method – based on the AprilTag affixed near the surgeon's grip – captures the actual hand-driven movement of the tool, accurately characterises the functional cutting behaviour during a controlled, linear cutting task and offering high-fidelity data on how motion intent translates into cutting performance. By decomposing the surgical trajectory into horizontal and vertical components, we provide a time-resolved view of the cutting dynamics that reflects real-world variability in device use technique. The BPR thus serves as a performance-linked metric that captures how each surgeon's motion strategy influenced effective vertical cutting progress. Overall, our method provides a robust, interpretable, and surgeon-specific assessment of bone cutting performance, grounded in both motion capture and force data. Future work may explore recording device rotations using the AprilTag, from which it would be possible to derive precise tip motion using the known tag-to-tip transformation matrix and the overall device geometry. Unlike purely theoretical or tool-tip-only tracking approaches, our system reflects the actual dynamics of surgeon-device interaction, offering superior insight into both technique and provide insight into device performance outcomes.

The results herein offer several useful insights. We confirmed that when surgeons use ultrasonic surgical devices to cut human cortical bone, it can significantly reduce intra-operative cutting force. The average measured cutting force across all surgeons and human cortical bone samples in this study was 1.91 N. These results are comparable with a previous study, where practising oral surgeons cut a fresh heifer (bovine) femur cubic sample with an approximate volume of

2 cm³ using an ultrasonic chisel, resulting in a similar orthogonal force between 1.48 N and 3.22 N.⁶³ Similar to studies conducted in highly controlled laboratory environments using mechanical test rigs instead of surgeons, deeper cuts can be achieved by applying a larger vertical force, however, the device will stall above a force limit, with the exact upper force limit dependent on the control system being able to track the changing resonance frequency under load and draw sufficient power to maintain the required ultrasonic displacement amplitude.⁷⁸ The overall trends in the motion capture data provided valuable insight into how the surgeons were manoeuvring the surgical device during use: deeper cuts can also be achieved by multiple horizontal passes at the same vertical force and going a bit deeper with each pass. In our study, S1 applied the lowest vertical applied force (0.65 N) across all samples and hence achieved the smallest cut depth (0.56 mm). Interestingly, there was also considerable variability in the way the surgical device was manoeuvred during cutting of a sample. For example, S2 applied the largest vertical force and hence advanced the surgical device much faster into the bone compared to S1 and S3. When S2 struggled to advance the tool into the bone, the motion locus of the device was much more complex, involving more changing of the angle of attack about the two axes (evident in Figure 3, where the device exhibits a more complex motion locus, involving frequent changes in the angle of attack about both the *x*- and *y*-axes). These variations suggest that the cutting motion was not confined to a simple linear path but involved compound movements, contributing to variations in the effective cutting direction and potentially impacting force and penetration rate. As the cutting force and cutting speed results for sample 1 (s1) between S2 and S3 were more similar compared to S1, comparison between these samples reveal that this more complex approach to cutting did not result in a deeper cut; average cut depths for S2 were 0.71 mm and for S3 were 0.79 mm. This would reduce the total number of horizontal passes and increase the time that the device is in contact with the bone sample being cut, which would increase the temperature at the cut site. High temperature of the bone at the cut site from friction between the bone and the blade is generated during cutting is an important consideration for developing ultrasonic surgical devices.^{38,46,47,53,79–81} Thermal damage can lead to osteonecrosis causing injury to bone tissue and bone resorption, which can adversely affect bone healing potential,⁸² cause implant loosening and jeopardise implant stability.^{38,83} While temperature can be significantly reduced through the use of intra-operative irrigation,^{38,84} the adoption of a consistent and simple cutting action by the surgeon can prevent localised heating and reduce the risks of osteonecrosis.

The currently reported BPR of 0.02 mm/s is very likely too slow for practical implementation in large-scale

orthopaedic bone cutting procedures. Given a mean articular tibial plateau width for males of 82.48 mm,⁸⁵ our current surgeons would require at least 4124 s (68.73 min) to cut across the width using the device in this study. To begin to translate bench testing results into effective clinical use and ultimately improve surgical performance and patient outcomes, we must begin by understanding from our results how a given surgeon may be interacting with a device. The average force of S2.s1.c1 and S2.s1.c2 of 1.92 N is comparable to the 1.87 N for S3.s1.c1 and S3.s1.c2, however, the vertical speed for S2 was 4.34 mm/s, which was over four times that for S3 (0.90 mm/s). The stalling observed in the ultrasonic device during S2 trials is likely due to this higher vertical feed rate, which may have exceeded the device's ability to dynamically adjust excitation frequency and power in real-time. This highlights the importance of balancing applied force and vertical feed rate during ultrasonic cutting. While previous work using porcine bone suggested that a contact load of 150 g (1.5 N) resulted in the greatest cut depth,⁸⁶ that study used a different cutting configuration – one focused on vertical penetration rather than lateral cutting. As such, its relevance here is limited, but it does underscore the broader point that applied load has a strong influence on cutting performance. An optimal cutting strategy would aim to maintain the lowest effective applied vertical force necessary to achieve effective cutting, while operating within a feed rate that does not overwhelm the device's control system or generate excessive heat or mechanical damage. These exact limits and ranges are still under investigation, specifically for large bone studies.

To mitigate the risk of device overload and improve consistency, integration with robotic surgical platforms offers a clear advantage. These systems can be programmed to maintain an optimal and consistent applied force during cutting, reducing the risk of excessive tissue damage, transducer stalling or high cutting temperatures.

Given the clear value in measuring and understanding how the surgeons' hands move the cutting device relative to the bone sample, for example, the vertical and horizontal cutting speeds, future work should explore the value of the 'surgeon use path' under realistic surgical conditions for making a cut and look at how that influence different cutting parameters such as for example, optimal parameters, the effect of using different blade profiles on cutting parameters, identify how generalisable the results reported in this study are and investigate how results compare before and after implementing surgeon training on optimal parameters.

Conclusions

Ultrasonic cutting devices, which operate with low force, may reduce MSP and the physical burden during

surgery. This study presents the successful development of a novel and clinically realistic experimental setup for measuring surgeon-controlled parameters during ultrasonic cutting of fresh human cortical bone within a controlled yet clinically realistic environment. By capturing real-time data on cutting force, speed and tool motion from practicing orthopaedic surgeons, we offer insights into how different surgeons each interact with these devices differently. Consistent with previous studies, our findings confirm that ultrasonic devices enable low force cutting (1.91 N). However, our results highlight significant variability in surgeon-device interaction, including how much force each surgeon applied, and how the device was manoeuvred which can influence cutting efficiency, thermal effects, and overall surgical outcomes. Given the critical importance of surgeon-related factors that directly influence device performance and clinical outcomes, our results highlight the need to understand how a given surgeon may be interacting with a device. This can inform training and device optimisation strategies, help translate bench testing results into effective clinical use and ultimately improve surgical performance and patient outcomes. Our results also support the potential of integrating ultrasonic devices with robotic platforms to maintain controlled cutting parameters and enhance consistency. Future work should explore optimal parameters, the effect of using different blade profiles on cutting parameters, identify how generalisable the results reported in this study are and investigate how results compare before and after implementing surgeon training on optimal parameters.

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