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Design and evaluation of an active lubrication brake for a surgical drilling simulator

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Abstract— **Kinesthetic and vibrotactile interfaces are essential for a wide range of medical training applications through simulation. However, their design remains a challenge in many cases to meet all the constraints of integration, workspace and price. To address one of the main limitations of these interfaces, namely the actuator, we have studied the use of the active lubrication principle. This approach enables to design a passive actuator capable of providing both kinesthetic and vibrotactile feedback while remaining compact and low-cost. Two such actuators were integrated into a tangible maxillofacial training replica reaching forces up to 4.5 N for the simulation of drilling procedures. To assess the utility of providing both types of feedback and the applicability of this actuator for surgery training, a study with fourteen non-surgeon participants followed by one with six maxillofacial surgeons was conducted. The results demonstrated the potential use of such actuators for surgery simulators and highlighted that providing both types of feedback was the best compromise between realism, user preference and performance.**

Keywords— Active Lubrication, Haptic Brake, Kinesthetic and Vibrotactile Feedback, Medical Training, Bone Drilling

I. Introduction

"Never the first time on the patient" is now a motto in many institutions and countries [1]; surgeons cannot continue to learn directly on the patient with mentorship. Thus, other learning methods allowing them to reach sufficient surgery skills need to be employed, as medical mistakes directly endanger the patient's life and body's integrity.

Traditionally, surgeons have learnt by practising on cadavers and animals. However, these training solutions raise many issues, such as ethics, limited availability, high cost, limited resemblance between animal or human cadavers and live surgery, and most of all, non-reusability in the long term [2]-[6]. To overcome this, technological solutions have been developed relying on the use of synthetic physical models (also referred to as tangible replicas) or simulators in virtual reality (VR) [2], as they aim to ensure countless training sessions, repeatability, flexibility and a certain level of realism [4].

These simulators and tangible replicas [2], [4]-[23] have been reported to help students be more confident, reduce their cognitive load and perform more quickly and accurately than those without any simulation training [2]-[5], [8]. When visibility is limited, like inside the patient's body, surgeons often rely solely on touch; haptic feedback is thus crucial. Kinesthetic feedback also enhances motor skill learning [3], [9], [24]. However, medical fields like the orthopedics lack haptic-based simulators offering both reusability and patient specificity (i.e. different characteristics for a same pathology), as well as realistic force feedback [4], [6], or, on the contrary, these fields have extremely realistic and reusable simulators that are very expensive and complex. This is also the case in maxillofacial training, a medical speciality focusing on the head and neck of the patient through reconstructive surgery, trauma surgery and plastic surgery.

Therefore, we present an adapted design of our previous work [41] and a more complete characterisation of a low-cost and compact haptic brake, which relies on active lubrication. Not only can it render different haptic textures with both vibrotactile and force feedback, but it can also be implemented as an embedded actuator inside a tangible replica or a virtual reality interface. Brakes have the advantage of being inherently stable and can only render resistive forces. Consequently, this actuator is a potential candidate for medical training tasks such as drilling bones, injection and biopsy, as these involve different haptic feedback from both the different tissues and the interaction with the surgical tool. We decided to evaluate this improved actuator for maxillofacial jawbone drilling training, as this operation requires high motor precision to avoid irreversible damages, such as the section of nerves [10], and involves lower forces than other drilling operations (e.g. up to 12 N compared

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to 200 N for tibia drilling [25]). As a first step, we used a tabletop design instead of a handheld tangible replica of a drill for this evaluation to facilitate the implementation of the experimental setup.

Currently there are no tangible replicas or simulators offering reusability as well as patient specificity for this maxillofacial operation while also being low-cost. Thus, our contributions are two-fold: the improved design of our initial brake, comprehensive characterisation and technical validation of a haptic brake, and the evaluation of its haptic feedback and of the relevance of having several types of haptic feedback in a training use case, presented in the following sections.

II. Related Work

A. Drilling training technologies

Undergraduate surgeons are often lacking necessary surgery skills and feel insufficiently trained [26], [27]: traditional methods like training on cadavers or animals pose serious ethics, cost, availability and non-reusability issues as well as being of limited resemblance to live surgery [4], [6]. In addition, new emerging solutions that do not convey haptic feedback (e.g. web-based devices or VR simulators without haptic devices) show worse performance after training [9] than those providing haptic feedback. Students have even reported the necessity to combine these non-haptic solutions with training on devices where realistic haptic feedback is provided [27]. This led researchers to develop haptic-based solutions for medical drilling, either in the maxillofacial field or others (e.g. spine and limb surgery). These solutions should reproduce the sensations perceived, such as the change of bone layers perceived during these drillings. Thus, researchers have focused their efforts on developing either tangible replicas or VR simulators with haptic feedback to recreate them.

Tangible replicas are physical replicas that recreate the form factor or the properties of human body parts during an operation. They achieve this mimicry by relying on the properties of substitute materials [11]-[16] or embedded actuators [17]. They have the advantage of using real or realistic mock-ups of surgical tools and provide the closest tactile experience to reality [4]. Many surrogate materials for bone milling and drilling operations were created [11]-[16]. However, due to their intrinsic nature, replicas do not enable long-term reusability. Indeed, these replicas can be damaged during drilling. To tackle this issue, some tangible replicas rely on rapid prototyping like 3D models [15]. This method still requires several hours of preparation to create the physical model. Thus, tangible replicas that are customisable, durable and ready to use without preparation time still require techniques that are more flexible. The use of actuators embedded inside a replica was considered as an alternative solution. For example, Ha-Van et al. [17] used a motor and a vibrotactile actuator embedded inside an orthopedic mock up drill. However, maxillofacial drills are much smaller [28] and need a more compact setup than theirs. In addition, vibrations should change with bone properties [17]: their setup used two types of fixed vibrations for each bone layer, while our solution enables vibrations dependent on the user force and bone layer. Moreover, our solution simulated a thin bone while theirs simulated a thicker bone. Finally, no mock-ups with embedded actuators have been created nor evaluated for maxillofacial drilling training, which is achieved in this work.

Another way of overcoming the previous limitations can be to use VR haptic-based simulators. These simulators have been widely used for 20 years for dental surgery [7]. Nowadays, simulators for other drilling operations have also been created, details can be found in reviews [4], [6], [18] and developed systems [9]-[10], [19]-[23]. They cover common orthopedic drilling in the hip or in the tibia [9], [18], [20], [21] as well as specific and complex drilling for hand and maxillofacial surgery [10], [19], [22], [23]. Some of them rely on complex structures and on robotic systems to render several degrees of freedom and high-fidelity haptic feedback (e.g. [10]). Due to the high-fidelity actuators used, they tend to remain expensive. In addition, these simulators tend to be under-evaluated [4]. These two factors combined are an obvious barrier to their wide adoption [29]. Hence, the remaining simulators rely on less costly haptic devices, such as the Phantom or Touch X devices, which have been used in the academic field [9], [18]-[23] and even commercially, albeit in more expensive setups [30]. These devices reach forces up to 7 to 12 N and have workspaces between 10 and 30 cm^3. However, these devices tend to have limitations in terms of freedom of movement and texture rendering that result in feedback less realistic than with tangible replicas or cadavers [6], [19], [21]. Furthermore, not all of them are intended to render vibrations and thus cannot accurately simulate vibrating tools [18]. These limitations could be partly solved by a tangible replica augmented with haptic actuators, as proposed in Ha-Van et al.'s work [17] or in this paper.

B. Kinesthetic actuators

Actuators for kinesthetic feedback have generally been divided into two categories: active and passive [31]. Existing simulators or mock-ups tend to rely on devices using active actuation [9]-[10], [17]-[23], [30]. By contrast, we decided to explore the use of passive actuation, not yet investigated. This type of actuation can only resist movement or force [31], and is inherently stable. Thus, it is a safe option that does not require the same complexity of mechatronics and control design.

Brakes and clutches compose the majority of passive actuation [31]. Other solutions like jamming exist [31]. Yet, they are louder and have a slow update rate, and hence are not adapted for real-time medical simulators. It is possible to classify the remaining solutions into two categories: non-modulated and modulated braking force actuators. Non-modulated braking force actuators have two states, blocked or loose. These brakes are easy to control, small, lightweight and low-cost [32]. However, they fail to render the stiffness and firmness of simulated surgical operations and tissues due to the lack of resistance modulation during movement. Thus, modulated braking force actuators are more suited to medical simulators.

Common systems include rubber pads [33], electrostatic brakes [34]-[35], eddy current brakes [36], disks [37], magnetorheological (MR) or electrorheological (ER) brakes [38]-[39] and active lubrication brakes [40]-[41]. The rubber pads are very cheap, easy to use and to control but are highly exposed to wear: their performance will decrease over time, preventing repeatability. Electrostatic brakes are small (less than

1.1 mm thick [33]), lightweight (51 g [34]) and provide high force feedback but use high voltage [33]-[34] (up to 5 kV [33]) and could be dangerous. Disk and eddy current brakes are also easy to control, yet they are usually bulky (i.e., the size of a palm [36][37]) and cannot be as easily implemented into haptic devices as other brakes [36]. MR/ER brakes can render a large range of forces and reach high forces, but can be difficult to design, as the fluid has to be confined in a sealed piece. Finally, brakes based on active lubrication [40]-[41] are low voltage (up to 37 Vpp [40]), low-cost, small (about 11 cm^3 [41]) and can generate high braking forces (up to 23 N [40]). Moreover, some of them can also render vibrotactile feedback [41], as demonstrated in our previous work. Thus, as drilling generates vibrations, they could be suitable for a medical simulator and allow the use of fewer haptic actuators than other existing mock-ups [17]. However, the brake presented in our previous work [41] relies on a manually adjusted screw for the compression, leading to potential repeatability issues. In addition, it was only characterised for its object grasping use case; the braking principle was not tested for other compression values nor the compression used was characterised. Therefore, our contribution is an improved brake design and a comprehensive characterisation of the principle of active lubrication for a haptic brake. We also assessed its integration and application into a tangible replica to provide a novel surgical training simulator able to render both kinesthetic and vibrotactile feedback while ensuring safety, compactness and cost effectiveness.

III. Design of the Tangible Simulator

A. Active lubrication phenomenon

Brakes based on active lubrication operate similarly to other friction brakes and can modulate the braking force by modulating the friction force. However, active lubrication brakes do not increase the friction to allow the modulation of force but instead reduce the existing friction to enable lower braking force [41]. To illustrate this phenomenon, we will briefly introduce how longitudinal vibrations help reduce the friction force. We decided to use longitudinal vibrations in this brake, similarly to [41], due to their low energy consumption and higher friction reduction compared to other directions of vibrations, namely normal and transversal [42]-[44].

We chose to use the Coulomb model because, despite being less accurate than other models [45]-[46], it is simpler, and the logic of the friction reduction remains the same. The Coulomb model states that, if a solid S1 is sliding over a solid S2 with a relative velocity $\vec{v}$, the friction force between the two objects is:

$$\vec{F} = -\mu\|\vec{N}\|\frac{\vec{v}}{\|\vec{v}\|} \qquad (1)$$

Where $\vec{N}$ is the normal load of S1 against S2 (i.e. the compression force), and μ the coefficient of friction, which depends on the contact materials' properties. $\vec{F}$ is always in the opposite direction of the relative movement of S1 over S2 and its norm does not depend on the amplitude of $\vec{v}$. Active lubrication occurs when S2 is activated with an oscillatory motion of angular frequency ω with a velocity parallel to the sliding direction of S1 (see Fig. 1):

$$V_v\,(t) = V_v * sin(\omega t) \quad (2)$$

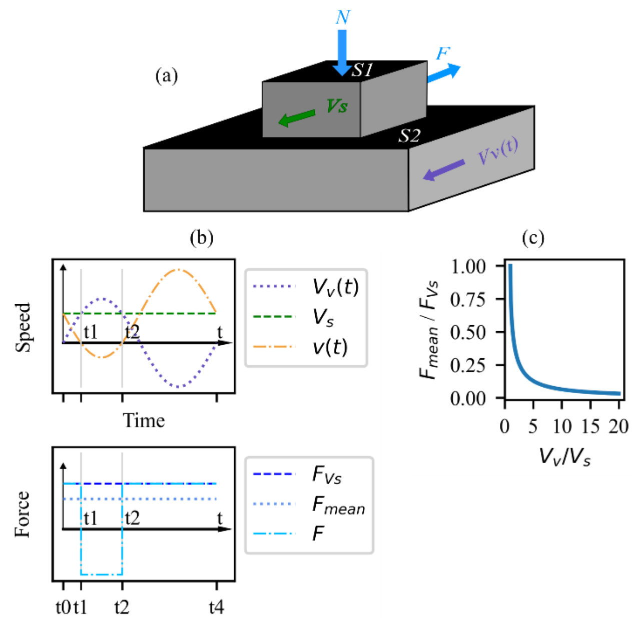

Fig. 1. Active lubrication in the longitudinal direction. (a) Schematic view of two solids S1 and S2, (b) Speed and Force of S1/S2 without vibration (V_s and F_{Vs}), and with vibration ($v(t) = V_s - V_v(t)$, $\vec{F}$ and F_{mean}). (c) Variation of the resulting friction force F_{mean} represented as a function of V_v/V_s, the ratio between the actuation amplitude and the speed of the displacement of S1 over S2. (Graph adapted from Kumar and Hutchings [42])

In the longitudinal direction, this motion, induced by a high frequency vibration, enables the reduction of the friction force perceived (i.e. $\vec{F}$). The velocity v becomes:

$$v(t) = V_s - V_v(t) \qquad (3)$$

Where V_s is the constant sliding speed of S1. The phenomenon is illustrated on Fig. 1, adapted from Kumar and Hutchings [42]. As illustrated in part b of Fig. 1, the resulting speed *v(t)* sign changes during a period of $V_v(t)$, leading to a change of sign in the friction force $\vec{F}$. Over a full period of $V_v(t)$, the friction force on S1 is then equivalent to an average force $\vec{F}_{mean}$ lower than the force $\vec{F}_{Vs}$, the friction force without the vibration actuation on S2. When used as a haptic device (when a user is moving S1 over S2), the change in direction of $\vec{F}$ is not perceived, as the actuation of S2 is usually in the ultrasonic region (above 20 kHz) and rather the user perceives a smooth overall reduction of friction ($\vec{F}_{mean}$). For the phenomenon to occur, the amplitude V_v must be greater than the velocity V_s. The resulting friction force $\vec{F}_{mean}$ is a function of V_v/V_s and is shown on graph (c) of Fig. 1: the higher the ratio V_v/V_s, the lower the friction perceived.

B. Brake design

We used the phenomenon of active lubrication to develop a brake for haptic devices. This brake is able to modulate the friction force. It is composed of a sliding plate (modelled by S1 in Fig. 1) compressed between two piezoelectric components (modelled by S2 in Fig. 1). The piezoelectric components create on both sides of the plate the vibration needed for the active lubrication phenomenon. Piezoelectric components used as actuators for active lubrication are a common choice [46]-[47]

as they are small, often low-cost and durable and they reach high frequencies.

We used the same setup as in our previous work [41] but improved it with a more repeatable compression mechanism. Piezoelectric buzzers (⌀15 mm, TDK PS1550L40N) were similarly operated at 193 kHz, a resonant mode where the vibration amplitude is maximum in the planar direction. The moving part of the brake (i.e. the sliding plate) was made of glass, as it is affordable and resistant to wear by friction. An amplitude modulation controls the piezoelectric buzzer: the modulation is the haptic signal desired and its carrier signal is a sinus at 193 kHz. Thus, the waveform and frequency of the haptic signal envelope correspond to the vibrations' parameters, while its amplitude affects both the braking force and the vibration amplitude.

After preliminary tests, we also decided to add jojoba oil onto the glass plate to limit the wear of the system and to increase its stability, as it helped the overall lubrication [41]. In fact, jojoba oil is low-cost and is known for its anti-wear, lubrication and high-pressure properties [48]-[49]. We have tested other lubricants, such as lithium and cork grease but the friction was not stable for the former (i.e. the friction increased in less than a minute) and the friction reduction was much lower for the latter (60 % at maximum). Exploratory tests were conducted to study the excitation frequency and the mechanical contact configuration: plates made of aluminium, steel, brass and piezo covered by tape were investigated. The sliding speed was not considered as a parameter even though studies [43], [50] have proven that it influences greatly the performance of active lubrication. This choice was justified by the fact that the relative speed is mostly influenced by the ratio between the speed of the motion and the maximal speed of the vibration. Indeed, at 193 kHz, the maximal speed is high (up to 200 mm/s), whereas the speed of the user in our use case is low in comparison (under 10 mm/s [10]). Thus, to simplify the exploratory tests, we chose to fix the sliding speed under 10 mm/s (around 7mm/s) because active lubrication allows higher friction reduction at low speeds [42], [45]. Nevertheless, in a future study, it will be necessary to determine the effect of speed on friction reduction to characterise fully the brake.

To characterise this new brake design's performance, we studied the influence of the piezoelectric buzzer excitation amplitude and the compression force on friction reduction. We used a test bench composed of an adjustable press above the top of the brake, a force sensor (Tedea Huntleigh compression load cell) under the brake and a 6-axis force sensor (ATI Mini40) mounted on a linear actuator. We define the percentage of friction reduction (FR) with F_{max} and F_{min} the maximal and minimal friction force reached, as:

$$FR = \frac{F_{max} - F_{min}}{F_{max}} * 100 \qquad (4)$$

We observed the same behaviour as in Dong and Dapino [43] and our previous work [41] studies. Indeed, the friction reduction increased to an asymptotic value with the increase of the voltage (see purple points on Fig. 2 (a)). Moreover, we observed slightly better performance and, as expected, higher friction force (Fig. 2 (b)) for higher loads (88 % for 20 N versus 73 % for 4 N). However, a compromise between the transparency (i.e. the residual braking force of the system when the target force is zero) and the maximum force reachable has to be found depending on the requirements of the use case.

Given these results, we decided to use values up to the maximum voltage available by the custom-made power card (i.e. 24 V peak to peak) to ensure the best possible friction reduction. To guarantee the values of the compression force and to allow disassembly for visual control of the wear of the plate between uses, the brake was set in compression by a torsion spring. The brake was 3D-printed in thermoplastic PC-ABS to limit the cost and the weight. A Raspberry Pi 0 and a custom power card ensured the signal modulation controlling the brake.

The brake developed measures only 4.5 cm * 2.5 cm *1.5 cm and weighs 7 g without the plate and 17 g with it (the plate weight depends on the length of the course needed) (see Fig. 3).

C. Tangible simulator design

As the aim was to evaluate the combined feedback of the brake on a simple medical gesture, we chose to create a low-cost medical module integrating it for maxillofacial drilling training. We chose to simulate the drilling phase of an Epker osteotomy (i.e. surgery to correct the jaw alignment of a patient), where the surgeon needs to drill precisely the jawbone into the Epker lines (see Fig. 4). This drilling phase then allows the surgeon to make a clear fracture of the bone and to move the jawbone in the desired position to correct the misalignment. A clean fracture is obtained when the cortical layer is drilled completely and no bone residues remain. In addition, surgeons must minimise the penetration in the spongy layer of the bone to avoid damaging the alveolar nerve in it [10]. In parallel, they must minimise the drilling duration to avoid thermal damage to the bone tissue [25].

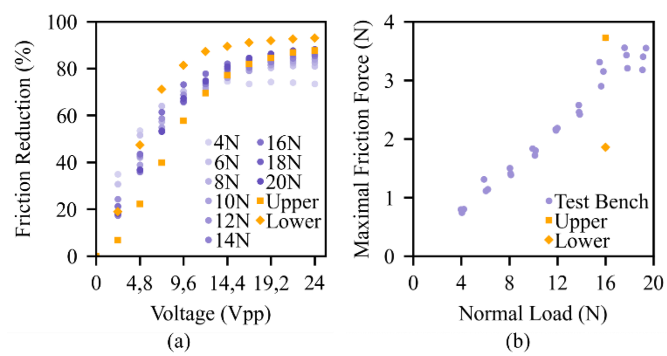

Fig. 2. Performance of the actuator: (a) Friction reduction as a function of the voltage amplitude peak to peak powering the piezoelectric buzzer at different compressions (b) Maximum force of the brake at 24 Vpp as a function of the compressions. Purple points: test bench measures, orange points: measurement from the brakes used in this study.

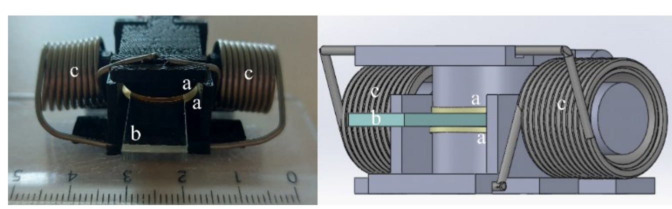

Fig. 3. Custom brake with a: piezoelectric buzzers (static parts), b: glass plate (moving parts), c: compression spring.

For a first version, we decided to simplify the training movement by focusing on a specific phase. In reality, surgeons have to place the drill carefully on the cutting line and ensure the verticality and trajectory of the drill (see Fig. 4). They repeat this drilling gesture over this line. Nonetheless, the most critical task is to stop drilling at the correct depth and feel the change of layers between cortical and spongy. Therefore, we focused on this step and specifically on this perception. In comparison to similar previous work but with a high-fidelity simulator, this is the first exercise out of three of Gosselin et al. [51], [10] focusing on vertical drilling.

The forces perceived during cadaver drilling in the vertical direction can go up to 12 N [10]. However, the average maximal forces applied by surgeons and measured in simulators during drilling were between 4 N and 6 N [10]. As the newly designed brake could only reach 3 N maximum (see orange point on Fig. 2 (b)), to reach such forces we decided to use two brakes compressed at 16 N (maximum force of 3 N * 2 brakes = 6 N). We chose to compress the brakes at 16 N and no higher because the spring achieving higher compression was too bulky. The resulting module is shown in Fig. 5. Due to manual assembly and its variability, the compression made by the springs was not in reality of 16 N, yet the overall friction force generated reached around 6 N (see Fig. 2 (b), points for upper and lower brake). The brakes were able to reach 89% of friction reduction, and forces between 0 N and 4.5 N after subtracting the weight of 1.3 N of the drill. This reduction pattern is similar to the reduction obtained during the exploratory characterisation tests (see Fig. 2).

The training module is a replica of a drill handle (Fig. 5, element b) glued to the moving part of the actuator (glass plate as element c), itself sliding between the two active lubrication brakes (d and e), as described in section II.A. In addition, a load cell (Tedea Huntleigh compression load cell) and a Linear Variable Differential Transformer (Sensorex LVDT SX 12N060) were used and read by a National Instrument card NI6211. They enabled to create a control loop on the vibrotactile texture applied to the brakes depending on the position and the force of the simulated drill. Indeed, the position in a layer needs to be monitored in order to convey the corresponding haptic rendering (typically cortical or spongy) as well as the force applied to match the corresponding drilling vibration [10]. For example, in the cortical layer, the force resistance applied on the drill by the brake and the frequency of vibration linked to the cutting head speed are higher than in the spongy one. Moreover, the speed of the drill's cutting head also changes with the force applied (see (5) and (6)). However, the braking force output of each brake was not monitored (i.e. open loop) and only the resulting force was, meaning that adjustment of the braking force during a same layer was not implemented. Two power supplies (Metrix AX0503A) powered the overall system.

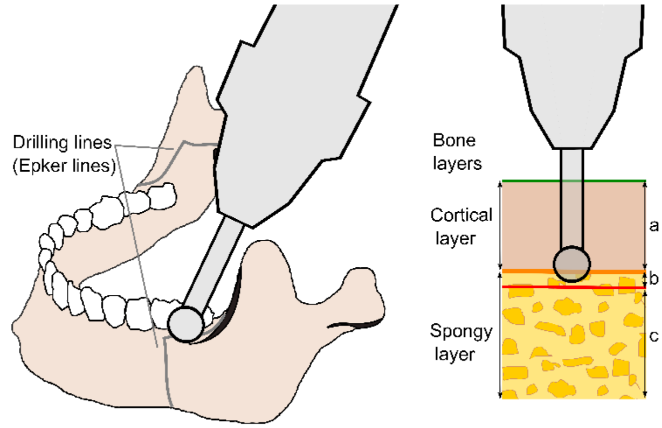

Fig. 4. Schematic representation of the drilling task, with a: cortical sensation, b: transitory sensation, c: spongy sensation.

The simulated drilling task has two implemented layers: cortical and spongy (see Fig. 4). However, we decided to render not only a sensation for each of the bone layers but also an additional one on the interface between the cortical and spongy layers. This additional sensation simulates the state where the drilling head is in between the two layers. The drilling goes through three successive sensations when going through two bone layers. First, the drilling begins in the cortical layer with the first haptic sensation simulating the cortical layer drilling sensation (zone a in Fig. 4). Then, when the drill enters the transitory phase between the cortical and spongy layer, the second haptic sensation is played (zone b) and is about half-diameter of the ball cutting head. Finally, as the drill head exits totally the cortical layer to be exclusively in the spongy layer, the third haptic sensation corresponding to drilling inside a spongy layer is rendered (zone c). The transitory phase was added to match the data collected from three surgeons when drilling on cadavers [10], where two successive drops in stiffness are perceived after exiting the cortical layer.

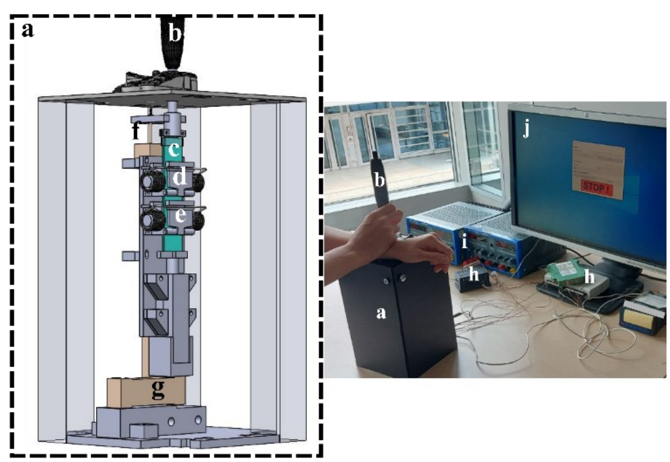

Fig. 5. Medical module and setup for the user evaluation. Left: 3D view of the simulator, Right: real simulator used by a participant, with a: Medical module, b: drill, c: glass plate, d: upper brake, e: lower brake, f: position sensor, g: force sensor, h: control cards, i: power supplies, j: screen with instructions for the participant

The overall system latency was 0.1 s due to the reading and processing of the position sensor values, while the rest of the setup operates at 5 ms. This latency did not create a mismatch in the texture but only a delay between the real position and the trigger of the new haptic texture. Thus, the depth of each layer was variable: this variation was about 0.1 s multiplied by the user speed. This is not an issue, as the depth of these layers varies from a person to another and with the localisation in the jaw [52] at a scale of about 1.1 mm. Hence, the transition layer depth stayed realistic for a speed under 11 mm/s. Since the task was about precision, we assumed such high speed would not occur (as confirmed by the study results, see section III).

As the brake is able to render force and vibration independently but also simultaneously, we deemed it interesting to evaluate the addition of vibration in the simulator and hence three separate conditions: kinesthetic and vibrotactile (KV), kinesthetic only (K) and vibrotactile only (V). The force applied by the brakes depends on the mean peak to peak voltage, as highlighted in Fig. 2 and in section II.A. The vibration generated by the brakes depends on multiple parameters and was a combination of two sinusoids (a fundamental frequency f and its second harmonic $2 * f$). The voltage difference between the maximal and minimal voltage determined the total amplitude of the signal. The ratio r provided the proportion of the total amplitude granted to each sinusoid. The frequencies of the sinusoids decreased with an increase of the applied force F as in [10]. These parameters are reported in Table 1 and (5), and the behaviour of the system matched the one of Gosselin et al. [10], which was validated by surgeons. We chose to set the fundamental frequency at 200 Hz. Indeed, we need to fit the device's bandwidth and enable a similar frequency decrease with the applied force as in [10] (see (6), where the initial period is 5 ms, 1.3 N is linked to the handle's weight and 0.5 is the decrease factor). The signal equation is shown in (5).

$$s(t) = [r \sin(2\pi f) + (1-r) sin(4\pi f)](V_{max} - V_{min}) + V_{min} \qquad (5)$$

$$f = \frac{1}{(5+0.5*(F(t)-1.3))*0.001} \qquad (6)$$

Table 1: Haptic Textures for Kinesthetic and Vibrotactile (KV), Kinesthetic only (K), Vibrotactile only (V)

		Cortical	Transition	Spongy
KV	***Mean voltage amplitude***	3.6 V	7.2 V	14.4 V
	Braking force	4.4 N	2.8 N	0.6 N
	Min-max voltage gap	2.4 V	3.84 V	4.8 V
	Ratio	0.6	0.6	0.8
K	***Mean voltage amplitude***	3.6 V	7.2 V	14.4 V
	Braking force	4.4 N	2.8 N	0.6 N
	Min-max voltage gap	0 V	0 V	0 V
	Ratio	0.6	0.6	0.8
V	***Mean voltage amplitude***	3.6 V	3.6 V	3.6 V
	Braking force	4.4 N	4.4 N	4.4 N
	Min-max voltage gap	2.4 V	3.84 V	4.8 V
	Ratio	0.6	0.6	0.8
	Simulated Force Applied and matching frequency f	4.2 N 155 Hz	4.2 N 155 Hz	0.5 N 217 Hz

IV. Evaluation on a drilling Use-Case

A user study was conducted to investigate both the relevance and usability of the feedback created by the passive actuator for a medical use case. First, we wanted to assess the utility and impact of having access to different haptic renderings on performance and realism. Second, we wanted to assess if the haptic feedback generated by the brakes embedded in the medical module provided, or could provide with fine-tuning, the desired haptic feedback of a surgery-training simulator. For this, the study was divided into two parts: first, an evaluation with non-medical participants, described in section IV.A, and second, an evaluation with expert surgeons, presented in section IV.B. The protocols used were mostly similar, with slight variations that will be described in the corresponding subsections.

A. Part 1: Evaluation with non-medical participants

In this first study, three haptic rendering conditions combining these modalities were compared: kinesthetic and vibrotactile (KV), kinesthetic only (K) and vibrotactile only (V). KV was chosen as jawbone surgery involves both kinesthetic, from the resistance of each tissue layer, and vibrotactile feedback, generated by the drill motor.

For K and V, previous studies have demonstrated the positive impact on training of either kinesthetic or vibrotactile feedback alone [24], [53]. In addition, when kinesthetic feedback was provided, adding vibrotactile feedback (without customisation dependent on the user force nor the bone layer) proved to increase realism but not training performance [17].

Thus, we deemed it interesting to integrate these conditions for comparison with KV as the comparison of K and V was not yet investigated and the vibrotactile feedback from other setups did not match reality. The goal was to assess the utility of each modality and its impact on performance and realism, as well as to collect some feedback about the user experience and other potential applications.

Regarding the haptic modalities, we hypothesised that:

H1. The feeling of realism and the differentiation between the cortical and spongy layers will be better in condition KV.

Indeed, Okamura et al. [54] reported that having vibration helps for layer differentiation, and Ghasemloonia et al. [22] explained that vibration is a cue used to differentiate layers.

H2. The task of differentiating between the two layers will be more difficult in condition V.

Ha-van et al. [17] reported that participants focus more on kinesthetic cues (stiffness in their case) than vibrotactile cues (though not dependent on bone properties in their case) to identify the change of layer.

H3. The duration of the drilling will be shorter in the two conditions that include kinesthetic feedback (KV and K).

If kinesthetic feedback is the preferred cue to identify layer changes, participants should be more confident with this cue and probably faster.

H4. Accuracy will be higher for condition V then for KV.

As the spongy layer is much less resistant than the cortical one in conditions KV and K, we expect the participants to move deeper in these conditions than in V for the same reaction time. Moreover, Okamura et al. [54] reported slightly less penetration with KV than K in a puncture task.

H5. Participants will prefer condition KV.

In their study, Ha-Van et al. [17] showed that having both force and vibration increased the fidelity of the training.

1) Participants

Fourteen participants (7f, 7m) were recruited within our institution as well as outside. Their age ranged from 24 to 53 years (*Mean (M) = 34.65*, *Standard Deviation (SD) = 10.40*), all were right-handed. The participants had varied backgrounds, with their professions spanning researchers, engineers in data, in metrology or in safety, executive assistants or technicians. None of them reported having any haptic sensitivity issues. Two of them reported extensive experience with haptic technologies (e.g. research prototypes).

2) Protocol

The study protocol was approved by the data protection officer of our institution, as well as the internal ethics digital committee. Before starting the study, the participants were asked to sign a consent form and reminded about their GDPR rights [55], and then asked demographics and background questions.

The main study was composed of three blocks corresponding to the three conditions (KV, K and V) with a counterbalanced order between conditions. Each block was composed of three phases: a training phase, a testing phase and a questionnaire. The setup is illustrated in Fig. 5 and described in section III.C.

The training phase aimed to familiarise the participant with the different sensations in each condition. They could first experience them separately during drilling (i.e. cortical alone, then spongy alone) and then together during a realistic drilling (i.e. cortical, transition and finally spongy), with this sequence repeated twice. They were given the instruction to minimise the drilling penetration into the spongy bone as well as the overall duration. The participant was guided by a visual interface displaying “STOP!” (c.f. Left of Fig. 5) when they exited the transitory layer (i.e. red line for b-c on Fig. 4). This was only available for training and they were instructed to use it as a confirmation of their perception. During the testing phase, the participant was asked again to perform ten complete drilling trials but without any visual help and with the same instruction about depth penetration and duration. Finally, for the questionnaire, the participant replied on a continuous scale between 0 and 10 (from low to high), about the ease of differentiating the sensations, the pleasantness of the haptic sensations and a self-evaluation of the task’s success. This aimed to collect preliminary and instantaneous feedback for each condition.

At the end of the three blocks, the participants answered a final questionnaire. This questionnaire was similarly composed of questions on a continuous scale between 0 and 10, about the ease of differentiating the sensations, the pleasantness of the sensations, the preferred condition and the subjective realism of the task, with additional questions about other possible implementations and use cases. Besides the answers to the questionnaires, data about depth penetration, force exerted and speed were collected. The experiment lasted about an hour.

3) Statistical Analyses

We performed repeated measures ANOVA for all normally distributed data (i.e. force in the cortical layer, self-evaluation and pleasantness) and Friedman ANOVA for the other (i.e. mean force, speed increase, accuracy, duration, ease of differentiating of textures and realism). We used Wilcoxon signed-rank post hoc tests with the Bonferroni correction for post hoc tests after a Friedman ANOVA.

4) Performance Results

Before analysing the variables, we verified that the setup was robust in the forces delivered. For that, we measured the forces across conditions in the first layer. Despite a small dispersion between measurements, the force during the cortical sensation was similar for all conditions: $F(2,26)=2,14$, $p>0.05$ (with $M_{KV}=3.1$ N, $SD_{KV}=0.4$; $M_K=3.23$ N, $SD_K=0.3$; $M_V=3.11$ N, $SD_V=0.3$), demonstrating a robust behaviour of the training simulator. There was also no significant difference in mean force between conditions, $X^2(2)=2.14$, $p>0.05$. Moreover, in conditions where kinesthetic feedback was involved, we expected a notable change in force for the beginning of both the transition and the spongy layers. Indeed, both these layers are softer than the cortical one and are simulated with smaller braking forces (see the higher values of the mean voltage in Table I). As depicted on Fig. 6 (cf. graphs for KV and K), a drop of force indeed occurred when the transition and the spongy layers began. In addition, when the transition layer began, a speed increase happened because this layer generated less resistance than the cortical one.

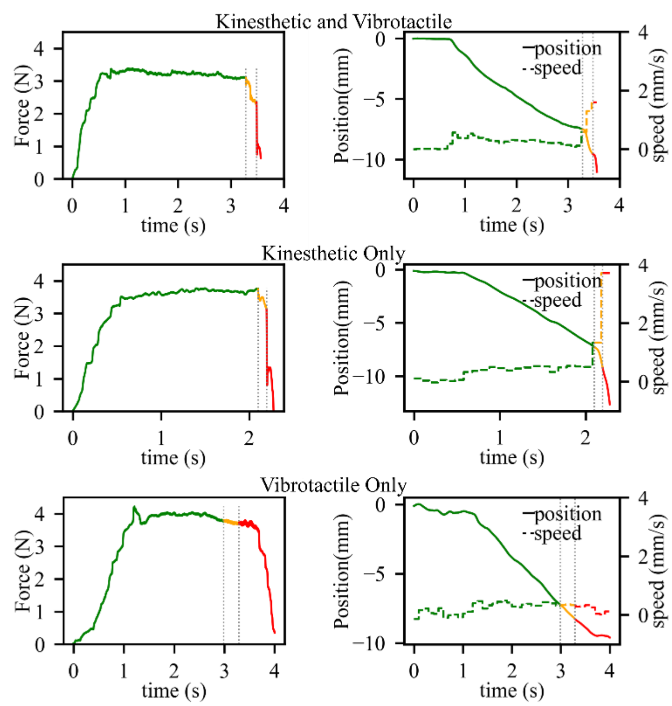

Fig. 6. Typical force and position of the tool during a drilling curve extracted from a participant trial. Green: cortical, orange: transitory and red: spongy sensation.

In the condition without kinesthetic feedback, i.e. vibrotactile feedback only (V), drops in force and an increase in speed were not present because the brakes did not decrease their braking force with the change of sensation.

To assess this, a factor of speed increase between the end of the cortical layer ($V_{CorticalEnd}$) and the beginning of the spongy layer (V_{Spongy}) is calculated as:

$$factor = \frac{V_{CorticalEnd} - V_{Spongy}}{V_{CorticalEnd}} \qquad (7)$$

There was a significant difference in the speed increase between the end of the cortical layer and the beginning of the spongy layer between conditions, $X^2(2)=21.57$, $p<<0.05$ ($M_{KV}=7.16$, $SD_{KV}=4.9$; $M_K=5.12$, $SD_K=2.7$; $M_V=0.92$, $SD_V=0.8$). Post hoc tests attest to a significant difference between KV and V and between K and V. For most participants in all haptic conditions, the strategy seemed to remain the same. They drilled at a null stiffness and tended to maintain force and speed as constant as possible. However, two participants made discontinuous drillings to better feel the haptic layer and control their movement.

For each drilling, two additional performance indicators were extracted: the accuracy (i.e. the distance between the beginning of the spongy layer and the minimal position reached) and its duration. We found a significant difference in accuracy between conditions, $X^2(2)=9.71$, $p<0.05$ ($M_{KV}=3.49$, $SD_{KV}=2.4$; $M_K=5.09$, $SD_K=3.9$; $M_V=2.09$, $SD_V=2$). Post hoc tests attest to a significant difference between KV and K and between K and V (see Fig. 7 (a)). This result partially validated **H4**, i.e. with the accuracy higher for V then KV.

There was also a significant difference in duration between conditions, $X^2(2)=9.14$, $p<0.05$ ($M_{KV}=8.63$, $SD_{KV}=7.8$; $M_K=5.54$, $SD_K=1.1$ and $M_V=8.35$, $SD_V=5.5$). Post hoc tests attest to a significant difference between KV and K and between K and V (see Fig. 7 (b)). This result invalidated **H3**, where we hypothesised that the duration of the drilling would be shorter in the two cases that include kinesthetic feedback (i.e. KV and K).

5) Questionnaire Results

The questionnaire at the end of each block was used to collect preliminary feedback. The ratings of the self-evaluations ($M_{KV}=6.86$, $M_K=6.39$ and $M_V=5.14$) and the pleasantness of textures ($M_{KV}=6.57$, $M_K=6.46$ and $M_V=6.25$ for the cortical layer and $M_{KV}=6.82$, $M_K=6.00$ and $M_V=5.71$ for the spongy layer) showed no significant differences between conditions ($F(2,26)=3,186$, $p>0.05$ and $F(2,26)=2,98$, $p>0.05$). The cortical layer and the spongy layer were reported as pleasant or neutral (i.e. score above or equal to 5/10) by 11 out of 14 participants. In addition, the ease of differentiation of textures was significantly better for conditions involving kinesthetic feedback. As this trend is identical to the one observed in the final questionnaire, we decided to focus the analyses on the final questionnaire, which accounts for the overall experience.

There was a significant difference in the ease of differentiating textures between conditions $X^2(2)=19.98$, $p<<0.05$ ($M_{KV}=8.46$, $M_K=8.50$ and $M_V=5.21$; $SD_{KV}=1.7$, $SD_K=1.5$, $SD_V=1.3$). Post hoc tests attest to a significant difference between KV and V and between K and V (see Fig. 8). Thus, H2 is validated, i.e. that discrimination was indeed more difficult for V and H1 is partially refuted, i.e. that KV was not more highly rated in terms of layer discrimination vs. K and V, just vs. V.

There was a significant difference in subjective realism between conditions $X^2(2)=9.14$, $p<0.05$ ($M_{KV}=7.89$, $M_K=5.89$ and $M_V=5.1$; $SD_{KV}=1.9$, $SD_K=2.9$, $SD_V=1.8$). Post hoc tests attest to the difference between KV and V (see Fig. 8). Thus, KV was perceived as the most realistic condition far before K and V that were rated similarly, partially validating H1.

Regarding the pleasantness, there were no significant differences between conditions, $F(2,26)=2;98$, $p>0.05$ ($M_{KV}=7.61$, $M_K=7.68$ and $M_V=6.11$; $SD_{KV}=2$, $SD_K=2.6$, $SD_V=2.9$). Overall participants reported that the textures were rather pleasant, i.e. with average ratings above 6.

Overall, the combined condition KV was preferred (ranked as first for 11 out of 14 participants) and condition V was the least appreciated (ranked as third by 9 participants). Thus, H5 where we hypothesised that participants would prefer KV was validated.

The final question probed the participants about other possible applications of the technology. They elicited an interest in other fields, such as the entertainment domain, and more specifically for video games, with 8 participants suggesting usage for a mouse click and 12 in joysticks.

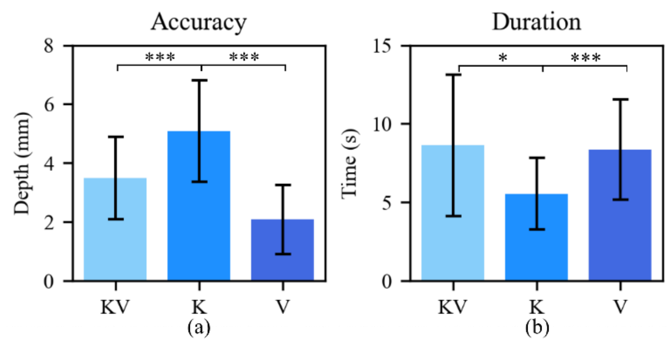

Fig. 7. Accuracy and duration

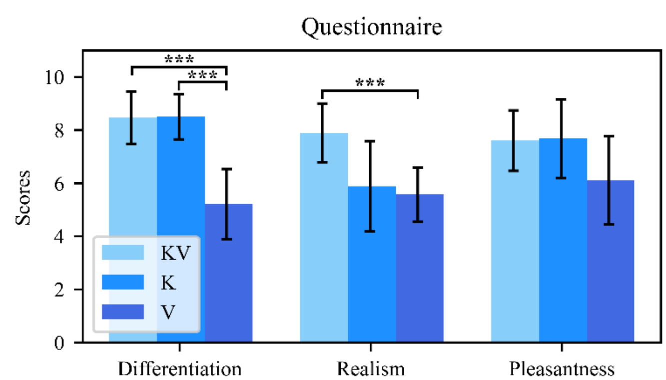

Fig. 8. Questionnaire average scores for each condition.

6) Discussion

Out of the five hypotheses made in the user study from section IV.A, two were validated, two partially validated and one refuted.

The first part of **H1** on the feeling of subjective realism was validated, as KV obtained the highest ratings. This can be explained by the richness of the haptic feedback and the adequacy to the expected feedback (based on experiences in drilling), involving both a change of resistance when drilling between layers and a change of vibrations generated by a drill depending on the user's force and bone layer. Even if our participants had no prior experience in cortical drilling, in the background questionnaire, all participants reported having experience of drilling in inhomogeneous material (occurring in home repair and DIY for 12 out of 14 participants) or at least mixing food.

The ease of differentiation (**H2**) was validated, and the latter part of **H1** was refuted. KV was better rated than V, but not than K. This result aligns with Ha-Van et al. [17], who found no difference in performance between K and KV, but contrasts with other studies [22], [54], where adding vibrations helped the task. A possible explanation is that the kinesthetic change was more perceptible than the vibrotactile change between textures. In fact, the resistance reduction between layers was about 1.8 N on average (SD = 0.7 for K and KV). The difference is well above the JND of 10^{-2} N [56]. However, the frequency change of the vibration was about 50 Hz and is just above the JND (i.e. 18% of a chosen frequency for a sinusoidal signal, i.e. between 24-34 Hz for the frequencies used in this setup [57]), and we have not characterised the amplitude of the vibration at the handle. Thus, we can assume that the frequency difference was perceptible, but we cannot account for the amplitude change. This would also explain why condition V was the worst in terms of ease of differentiation, with more subtle differences, and it possibly added difficulty in condition KV.

The hypothesis on the duration of the drilling (**H3**) was partially validated and partially refuted. Indeed, even if the duration was significantly shorter for K, it was similar between KV and V. One of the possible explanations is the difficulty of perceiving the layer change through vibrations, as highlighted in the previous paragraph. This affected both V and KV, by trying either to keep the speed slow and constant as in V or on the contrary to slower it as in KV to help focus on the discrimination.

The hypothesis **H4** on the accuracy was almost validated but not in totality. Indeed, as in the study of Okamura et al. [54], the accuracy was better for KV than K. We even had significant differences, most likely as in our case, there were no vibrations occurring at the change of the haptic layer in the condition K thanks to our actuator choice (brake in our setup vs. motor in theirs). However, there was no significant difference between V and KV. Initially, the hypothesis was formulated based on the impact of the braking force on controlling the speed increase when changing layers, assuming a similar speed increase for both KV and K that would impact the accuracy. For V, as the speed can be easily kept constant, the task is primarily about discriminating the textures. For K and KV, the resistance felt imposes the need to adjust the speed. Yet, as depicted on the right of Fig. 6, we can note that K has a higher average speed than KV, which, in turn, implies more depth travelled before reacting and stopping, and underlines an impact of vibrations on the speed in KV.

User preference validated **H5**, as condition KV was largely preferred over other conditions. This can be explained by the fact that the participants deemed this condition more realistic, easier in terms of differentiation and relatively more pleasant in terms of haptic textures. Once again, like for **H2**, this finding is in line with the results of Ha-Van et al. [17].

B. Part 2: User study with maxillofacial surgeons

After validating the prototype and recognition of its feedback with non-medical participants, we conducted a pilot study with expert maxillofacial surgeons to validate the applicability of this new design of the brake for jawbone surgery training. The goal was two-fold, validate the realism and preference of the feedback generated with experts and collect inputs on how to design a realistic simulator using the feedback created by our brakes. In light of the results of the first user study, where the vibrotactile condition led to more difficult discrimination (**H2)**, was worse in duration than K (**H3**), deemed lest realistic (**H1**) and finally least preferred, and due to the busy schedule of expert surgeons only two haptic rendering conditions were compared: kinesthetic and vibrotactile (KV) and kinesthetic only (K). In addition, this removal was also validated with one of the authors' experiences in simulators for drilling cortical bones.

1) Participants

Six expert maxillofacial surgeons (3f, 3m) were recruited at *the Amiens hospital (France)*. Their age ranged from 32 to 77 (*M=51.17, SD=14.27*). They had experience in surgery from 8 to 51 years (*M=25.67, SD=15*.07) and practised drilling surgery at least once a week. They were all right-handed. None of them reported haptic sensitivity issues. Five of them reported having experience with haptic devices for medical training from research projects. Their participation was voluntary.

2) Protocol

The study protocol and setup were largely similar to the one described in section IV.A and was also approved by the data protection officer, as well as the internal ethics digital committee. Before starting the study, the participants were asked to sign a consent form and reminded about their GDPR rights [55], and then asked a few demographics and background questions.

The main differences with the previous study were that this study was composed of two blocks and not three (i.e. conditions KV and K) but also with a counterbalanced order between conditions, a single questionnaire administered at the end of the study and an optional additional exploratory design phase. Each block was composed of the same two phases (i.e. a training and a testing phase). The instruction remained the same with the minimisation of the penetration into the spongy bone and the duration of the drilling, with the same number of trials in the training and testing phases as in the first study (i.e. 2 for training and 10 for the testing). At the end, for the final questionnaire, the participants were asked questions on a continuous scale between 0 and 10 about the ease of differentiating the sensations, the realism of each layer but this time solely for the most realistic condition, and the realism of the task in general. In addition, they were inquired about possible improvements and other foreseen medical use cases. During the optional exploratory design phase, surgeons were asked how textures could be changed to improve the realism. If these changes pertained to modifiable parameters of the setup, such as the force resistance, vibration amplitude, frequency or signal shape, then these were tested and surgeons asked for feedback.

3) Performance Results

Due to the limited number of participants, statistical analyses were not conducted, and thus sole tendencies are presented. Concerning the duration and the precision of the drilling, the surgeons were able to complete drilling in less than 5 s on average (i.e. $M_{KV}=3.24$, $SD_{KV}=2.38$; $M_K=2.73$, $SD_K=2.03$) and to drill less than 4 mm into the spongy layer on average (i.e. $M_{KV}=4.26$, $SD_{KV}=1.71$; $M_K=3.63$, $SD_K=1.89$). Differences in duration and accuracy between conditions K and KV were minor. Indeed, surgeons explained that in reality, they control the vibration speed with a pedal; consequently, they do not rely on the vibrotactile feedback to detect the layer change. We can assume that they focused only on the kinesthetic change.

4) Questionnaire Results

The ease of differentiating between the cortical and the spongy textures was rather highly rated by surgeons, with scores of about 8.5 for each condition (i.e. $M_{KV}=8.58$, $SD_{KV}=0.84$; $MK=8.5$, $SD_K=1.10$). One surgeon even commented during the training that the message "stop" was irrelevant to them. This underlines that the prototype and feedback managed to reproduce the fidelity of the task to some extent. However, two participants found the transitory layer too long and too easily felt, and thus it could be confused with the spongy layer. In fact, in their experience, during surgery, they only feel the cortical and the spongy layers and there is no additional transition sensation This contrasts with measurements of this phase made by Gosselin et al.'s [10]. Five out of the six participants rated KV and K similarly, most likely as they do not rely on vibrations during real surgery to discriminate the layers, and rather solely focus on the kinesthetic feedback.

The realism of the gesture was rated higher for KV ($M_{KV}=7.53$, $SD_{KV}=0.67$) than for K ($M_K=5.36$, $SD_K=2.42$). Indeed, similar to the first study, KV was rated higher as it corresponded more to the real behaviour, combining kinesthetic feedback from the change of layers and vibrations from the drill itself. One surgeon employed a similar metaphor as a participant in the first study to describe the kinesthetic only condition as not realistic at all by stating that it felt *"like a knife in a cake"*. Another surgeon even gave a score of 0 out of 10 because vibrations were deemed crucial. Interestingly, three of them found it disturbing to begin the drilling directly into the cortical layer and not in the air above the bone, as happens in reality. In any case, five surgeons preferred KV over K, with the remaining surgeon considering them as equally good.

The realism of the cortical layer was rated about 6.42 out of 10 ($SD=1.88$) and the realism of the spongy layer was about 4.83 ($SD=1.97$) in the combined condition KV. These low scores seem to contradict previous ratings, but testify of the margin of improvement of the haptic textures. The ratings were often associated with at least an inaccurate vibration amplitude or inaccurate layer resistance. Indeed, half of the surgeons reported that the vibration amplitude was too low in the cortical layer and one of them found the vibration amplitude as too high for a head drill, as simulated here. Following this feedback, as the parameters could be fine-tuned, a new texture with higher amplitude was proposed for testing. Surgeons who had previously found the vibration too low found it improved and more accurate with this new value. However, two surgeons reported that with more vibration, it felt more realistic but that the sensation fitted a twisted drill more than a head drill. Concerning the layer resistance, the reasons for the inaccuracy also varied depending on the surgeon. Two reported that the spongy layer lacked some resistance. On the contrary, another participant found the texture too strong. He was used to letting the drill sink by itself through its weight into the patient and only focused on holding the drill when the layer change happened. Given the variety of feedback and the contradictory suggestions, this warrants further investigation.

When inquired about other potential applications, they all reported that this system should not be limited to maxillofacial surgery training and should be expanded to other osteotomy operations (i.e. operations for bone deformation reparations) involving a change of layer, such as for a bi-cortical osteotomy in the leg. As a matter of fact, maxillofacial surgeons all reported that they have learned to differentiate and master the drilling through supervised osteotomy on other body parts, such as in orthopedics, on a real patient, before performing maxillofacial surgery. Thus, in their experience, they never had the opportunity to haptically train before operating for the first time and found the setup even more interesting because it could help future generations. In addition, the fact that the replica was ready to use and did not impose wearing equipment, such as for VR simulators, was strongly appreciated by one of the surgeons. Also, having one replica by operation phase or operation tool was not considered problematic.

V. Discussion

A. Applicability of active lubrication for a surgical simulator

The tangible replica presented in this paper is able to render both kinesthetic and vibrotactile feedback. The user studies assessed that our system was quite stable and that the combination of both types of feedback was greatly appreciated. This suggests that a brake based on active lubrication shows potential for applicability to a surgical simulator.

Yet, we observed on the test bench an impact of wear on the system's performance, and also partly between the two studies. Indeed, after a certain number of trials, we observed an increase in the braking force due to the tearing of the contact surface of the piezoelectric buzzer by the glass plate. For example, the average braking force in the cortical layer was about 0.75 N higher with surgeons than during the first study, and the dispersion between measures also increased ($SD=1.75$ instead of $SD=0.4$ for the first study). This wear can be managed in three ways: either by employing a more resistant material for the contact surface, by cleaning the brake regularly or by adding a control loop for the force that would continuously monitor it and adjust the parameter (e.g. voltage) for consistent feedback. However, the latter addition would also increase the complexity, the bulkiness and the price of the setup as it would require the integration of a sensor with more sensitivity and its control card. Thus, the optimal solution would depend on the use case requirements and the compromise between the criticality of precision and affordability.

Furthermore, the system currently uses a position control loop that is too slow to ensure high precision and real-time feedback. Indeed, the latency is about 0.1 s. The recommended feedback rate for haptics is at least 0.03 s and preferably 1 ms [56]. This drawback does not impact the results of the study presented here but can affect the realism of training where the depth of each layer needs to be precisely monitored. This can be easily circumvented with a more precise and faster sensor (e.g. hall sensor wit pulley-belts setup) as the rest of the setup is able to work at a 5 ms latency, though it could affect the cost.

B. Comparison with other drilling simulators

In the literature, the closest research to our work is the mock-up drill from Ha-Van et al. [17] designed for another and thicker bone drilling procedure and the costly and high-fidelity training platform by Gosselin et al. for the Epker surgery [10]. Even though our tangible replica does not render the exact same forces and texture patterns, it is interesting to compare condition K and KV to Ha-Van et al.'s results [17] as well as KV with Gosselin et al.'s results [10].

In the study of Ha-Van et al. [17], non-surgeon participants performed bone drilling in both K and KV conditions and were instructed to perform drilling through an entire bone (i.e. with three layers: cortical, spongy and cortical) and limit the distance travelled outside the last layer of the bone. The cortical resistance was set to 4.0 N/mm and 1.5 Ns/mm while the spongy one to 0.05 N/mm and 0.5 Ns/mm. Two vibrations were rendered; one for outside the bone and one for inside. Contrary to our results, they found no significant difference in accuracy between K and KV conditions. This is potentially due to the difference in the task; ours stopped in a layer with reduced but existing resistance, while theirs stopped outside a resistant layer, meaning that our participants spent less effort to hold back the drill, also leading to lower distances travelled (15.1 against 3.49 mm in ours for K and 15.53 against 5.9 mm in ours for K). Nonetheless, as in their study, the same tendencies in terms of user preference and realism were observed.

In the study of Gosselin et al. [10], the same task was performed by surgeons with similar background and experience than those in our study. Similar performance was reached with our less complex setup (i.e. one degree of freedom versus six degrees of freedom in Gosselin et al.). Indeed, our six surgeons completed the drillings in between 2.8 and 6.3 s, and on average in about 4.6 s, whereas their four expert surgeons completed their drillings in between 3 and 4 s. Our surgeons also performed the drillings with a depth between 0.5 and 7 mm and on average 3.2 mm, whereas their surgeons performed the drillings with a depth between 3 and 4 mm. In summary, our study obtained similar averages for the duration and depth of drilling for a similar depth of cortical bone, which is a promising result. However, higher dispersions were observed between our surgeons' performance, which can be explained partly by the mechanical differences of the systems and most likely by the small size of participants in both studies.

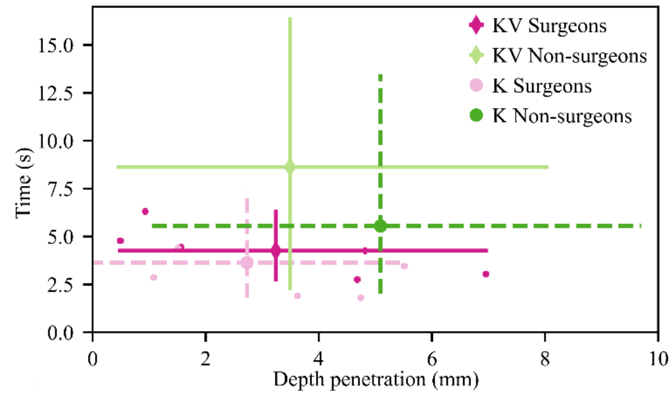

Fig. 9. Surgeons compared to naïve participants' performance in terms of duration and accuracy for the Kinesthetic and Vibrotactile (KV) and the Kinesthetic only (K) conditions.

Similar to the results of both studies, we can assume that the performance of novice participants would improve after weeks of training. Indeed, even if our first study involved non-surgeon participants and our second surgeons, surgeons were generally more precise and faster than participants from the first study (see Fig. 9) as they are familiar with this task, contrary to the naïve participants. Thus, a long-term study should be conducted to assess the effectiveness of our training module compared to other existing solutions.

C. Haptic sensations

The haptic textures displayed were deemed as realistic by non-experts but lacking realism in some cases according to surgeons, in particular for the spongy layer. This can be potentially overcome as the parameters, such as the resistive force, the signal shape, the frequencies and amplitude of the vibrations, can be fine-tuned.

Concerning the resistive force, surgeons asked for more rigidity in the spongy layer. This can be easily modified as the spongy layer force is obtained with the maximum friction reduction. Thus, conveying a higher force can be achieved in two ways, depending on the minimal force required. First, we can decrease the friction reduction by lowering the mean voltage assigned to this layer. Second, we can decrease the weight of the handle and consequently increase the force required by the user to move it.

Concerning the sensation of moving deeper into the layers, carried mostly by the vibration, one surgeon felt that it was not provided with the current signal, with a sinusoid shape. When he tried a square signal, he reported that this signal was more adequate for the cortical layer, as he felt the progression of the drilling more, and perceived the vibrations more strongly. This was not noted by the other surgeons, but changing the signal shape could be another lead to improve realism.

For the vibration amplitude, some surgeons found the vibrations not strong enough in the cortical layer and others found them too high in amplitude for a head drill, as simulated in the system. Thus, even if the resistive force and the amplitude of the vibrations are linked (i.e. the mean voltage controls the force and the min/max voltage gap controls the amplitude of vibration), some improvements can be made. Indeed, reducing the vibration amplitude is always possible by reducing the min/max voltage gap. However, increasing the amplitude of vibration can be more complex. The power card delivers a maximum of 24 V peak to peak; thus, the maximal amplitude is reached for a min/max gap of 24 V, which also sets the force. To overcome the limitations of this coupling, three solutions seem possible. First, if the amplitude vibration at high force is too small, more than two brakes could be implemented. Indeed, if there are more brakes, the same braking force is reached at a higher percentage of friction reduction that corresponds to a higher mean voltage amplitude and allows for a higher min/max voltage gap. Thus, the vibration amplitude could be increased without modifying the resulting force. Second, with a compression technique other than a spring, the brake could be compressed at a higher normal load and, in theory, reach higher maximal friction reduction (see (1)). Third, an actuator generating vibrations could be added to the replica. The first solution will lead to a bulkier and slightly more power consuming setup. The second solution could lead to a less robust system more susceptible to snap and wear, whereas the last would lead to a more complex setup.

VI. Future Work

As the objective of this paper's studies was to assess the utility and usability of the haptic feedback provided by our active lubrication brake, the replica design was simple. It did not allow the same freedom of movement, nor the realism of feedback expected in a handheld replica. Thus, an upgrade on the replica form factor as well as further studies on the expected haptic feedback should be realised before conducting user studies on the long-term training, as detailed below.

First, concerning the form factor, the actuators and probably some of the sensors should be moved from the case below the 3D printed jawbone to the drill handle interior. Thus, just like the mock-up of Ha-Van et al. [17], the drill tip would retract inside the drill body when the user tries to drill the jawbone, creating the movement of the drill toward the jawbone and the illusion of penetration. This upgrade would also allow the user to perform a drilling in any direction, enabling the same mock-up to simulate drilling along the Epker lines with only a more complete 3D printed skull. For this purpose, the actuator width should be slightly reduced before fitting it inside the handle of the drill and the sensors or sensing system should be changed. A lead to reduce the width of the actuators could be to change the compression method as well as change the width of the piezoelectric buzzers and the glass plate. For the sensor part, the force sensor should either be reduced and placed at the end of the drill retractable tip or stay inside the 3D-printed jawbone, while the position tracker should be changed for a smaller one, such as a system of pulley-belt with a hall sensor.

Second, concerning the drilling sequence, two upgrades should be implemented. Indeed, the surgeons had diverse opinions on vibration amplitude and frequency, as in reality they control the motor rotation of the drill with a pedal. To match their real experience, the vibration frequency could be user-chosen, either through software settings or through a simulated pedal. In addition, most of the surgeons reported being disturbed by the absence of a sensation of tapping on the cortical layer. Indeed, they were expecting, as in reality, to feel the drill rotating in the air before the first contact with the cortical layer. This could be implemented by including an additional layer without resistance and solely with vibrations before the cortical layer.

VII. Conclusion

This paper presented the improved design, characterisation and evaluation of a haptic brake based on the principle of active lubrication with ultrasonic vibration, which is able to generate both vibrations and resistive forces. As this actuator is low-cost, safe and compact, it can be easily integrated into haptic simulators. To evaluate its feedback in a realistic use case and because of the lack of haptic training simulators for maxillofacial surgery, it has been embedded into a tangible replica simulating jawbone drilling. The tangible replica achieved force feedback between 0 and 4.5 N as well as vibration amplitude and frequency modulation by using two brakes.

Two user studies were conducted, the first with 14 non-medical participants and the second with 6 expert maxillofacial surgeons. In these studies, different combinations of possible haptic feedback, i.e. kinesthetic and vibrotactile, were tested to assess their utility and impact on performance. The results showed that the condition with both haptic feedback was not the best in terms of accuracy or duration, compared to the kinesthetic only or vibrotactile only conditions, yet it achieved good performance, usually in between the two conditions or close to the best one. For surgeons, the performance was nearly identical to the kinesthetic only condition. The condition with both feedback was nonetheless generally preferred and deemed more realistic in particular by expert surgeons. The surgeons proposed several leads of improvement to ensure a higher realism of the sensations and of the setup and confirmed the need for such a simulator.

In light of these results, future work will focus on improving the haptic feedback, as well as the ergonomics of the setup. Once improved and validated, further studies will be conducted to assess the device as a standalone training simulator, with repeated sessions and impact on performance.

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