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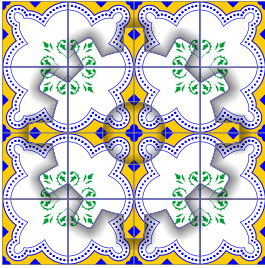
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EXTERNAL THREAD MILLING ON BOVINE CORTICAL BONE

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Abstract. *The most commonly used implants for orthopedic and orthodontic applications are the metallic ones, specially, those made of titanium. In spite of presenting high mechanical strength and remaining chemically inert, these implants are not bio-absorbable and are susceptible to biological rejection. An alternative that would eliminate such complications would be the use of bone screws. The purpose of this paper is to study the thread milling process on a bovine cortical bone insert. The experiment was performed on a CNC machining center, using a helical thread tool with multiple cutting flutes. The helical trajectory is constructed with several linear interpolations to be used in regular CNC machines without helical interpolation. The cutting forces involved in the process were measured using a dynamometer and the maximum forces in each flute is analyzed and presented in the article. Bending component force in the tool showed to be very low comparing to the resultant force. Forces Tangential to the thread surface have values around 50 N while resultant force achieved 57N. It is analysed the thread quality and cutting forces and it shows that the bovine cortical bone behavior during the machining process does not differ much from that presented by most common materials.*

Keywords: *Thread Milling, Bone Machining, Biomechanics, Dynamometry.*

1. INTRODUCTION

There are two different basic types of thread bones: cortical threads and spongy threads. Spongy threads have deep depth and large pitch, which increase the surface contact of the device with the bone and, consequently, its strength against pull off. Screws with this type of thread can have thread on all the length of the shank (fully threaded screws) or just on its tip (partially threaded screws). Cortical threads have less deep depth and smaller pitch than spongy thread. Moreover, they are fully threaded and, generally, blind end. This type of bone screw is also used to fix plates to the bone. In Figure 1, can be viewed the difference between the geometry of the two types of thread.

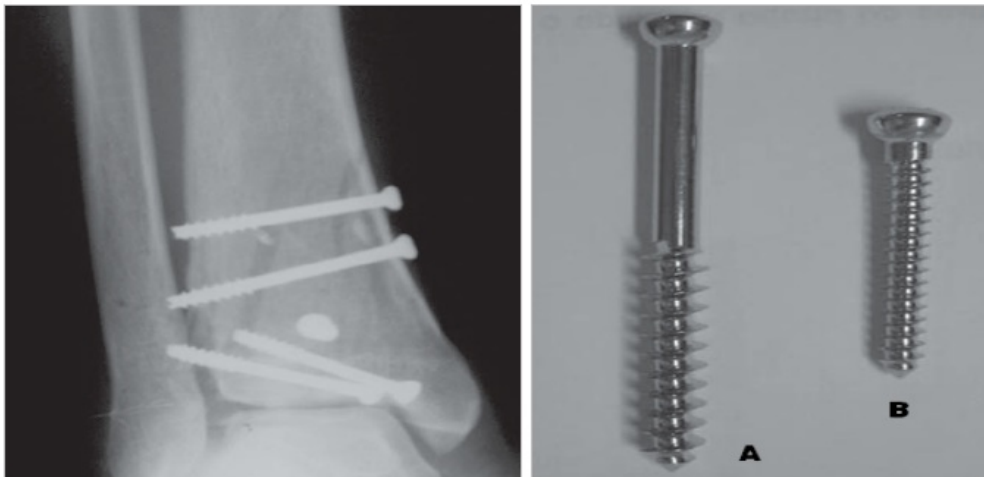


Figure 1: Metallic threads used on bones (Bretas *et al.*, 2009)

Currently, the most used implants on the treatment of a fracture are the metallic ones, specially, the ones made of titanium. Such material, besides having great mechanical resistance, it does not deteriorate and it stays chemically inert, that is, it does not react with other substances. Sometimes, it is also used screws manufactured from bioabsorbable polymers. These polymers are a type of plastic especially developed to allow the body to absorb the material after some time. The absorption of this material happens because it is identified by the organism as a dead bone tissue and needs to be substituted. Therewith, the polymeric screw is slowly absorbed and substituted for alive bone tissue. Although widely applied, metallic implants are not bio-absorbables and can cause a series of problems, such as: biological rejection; damage in the biomechanical performance in the repaired area; loss of bone mass near the implant; displacements in the bone-implant interface caused by the difference of rigidity; and the necessity of a second surgery to remove the screw after the consolidation of the fracture (Bento, 2003). This additional surgery can cause damages to the patient bone as the screw use to be wrapped by an amount of tissue that is formed around it as a defensive mechanism of the organism.

A lot of studies of bone implants used to odontologic applications verified that the overload in the bone-implant interface is a decisive factor in the bone loss that is observed around the implants. In the case of orthopedic implants to the repair of fractures, the bone adaptation to the mechanical load can result on an adverse redistribution of the stresses and, consequently, the bone reabsorption. This phenomenon is the result of the interaction of mechanical and biological processes and happens the following way: after the implantation, the prosthesis, having greater rigidity than the bone, protects it from the loads making the adjacent amount of bone to atrophy and be reabsorbed in response to the reduced load that it bears now. In extreme cases, the liberation of the implant and the increase of the incidence of fractures can happen (Oesler *et al.*, 2004).

In vivo studies have showed that a reduction of 1.2 to 1.6 times on the resistance to the extraction of the screw happens in the twelfth week after the implant placement, because of the bone reabsorption phenomenon (Neto *et al.*, 1999). Even the polymeric materials also have inconveniences: they can be toxic and can cause an inflammatory process in the region they were inserted, besides having an extremely high cost (Bento, 2003). A third alternative that would eliminate all these unwanted complications would be the utilization of screws made of bone. These devices, still in experimental phase, are capable of acting as structural element during the treatment and are, later, absorbed by the organism, avoiding, this way, a new surgical intervention to their removal. Additionally, when integrated to the receptor bone, the bone screw causes a more uniform distribution of the deformations and of the loads borne. If the incorporation is well succeeded, there is a reduction of the interface of prosthesis-bone contact with the treatment and the material becomes more homogeneous, once the dead bone - the prosthesis - is substituted for newly formed bone. As a result, there is a behavior that comes near of the one bones without implant have and the response of the bone-implant system to external loads becomes more homogeneous. Besides, the removal and deposition cellular activities of bone mass take the bone tissue to a constant remodeling, that varies according to the mechanical solicitations. In the implantation region of the bone screw, contrary to the metallic one, these cellular activities continue to happen and the implanted dead bone is substituted by a new alive bone.

However, it is important to highlight that the application of bone screws must be restrict to some specific conditions that do not demand high resistance like maxillofacial and periodontal surgeries. The bone to be utilized in the manufacturing of these implants does not need to be originated from humans. The bovine bone tissue, according to the results obtained in Bento (2003), is capable to induce the bone integration even when it is inserted in bones of other species. Moreover, bovine bone has superior structural properties in comparison to human bone (Cowin, 2001).

Considering the described technologies, thread milling process is adapted to the machining of cortical bovine bones. This technique allows dry cutting, cutting velocities that are not connected directly to feed velocity and good quality surface workpieces. Those characteristics are essential when a brittle material needs to be machined, as a bone which could not support high feed velocities as the thread pitch per revolution.

The objective of this article is to contribute to the manufacturing process of thread in bovine bone and the development of implants in biocompatible materials to medical and dentistic applications. The workpiece consists in a cortical part of bovine femur and, as per its reduced information of knowledge on machining proprieties, values from drilling bones were used. The machined parameters are shown and the experiment is described in the next sections.

2. BONE PROPERTIES

The bone tissue is a very rigid connective tissue found in the skeleton of the vertebrates. It is mainly functions are: support the organism, serving as a base for the body; allow the performing of movements; protect vital structures as the heart, lungs and brain; hold mineral salts; produce cellular elements of the blood; besides of serving as mechanical base for the movement. The bone composition, in weight, consists of 65% of mineral, 20% of organic matrix and cells and 15% of water Cowin (2001). The mineral part is constituted, basically, of hydroxyapatite (calcium phosphate) which, despite its rigidity and fragility, gives the bone enough resistance to compression, thanks to the way it is attached to the collagen. The bone tissue can be divided in two types: spongy (or trabecular) and compact (or cortical). The difference between these two forms is associated to their porosity and, consequently, their density.

The spongy bone tissue has larger medullar spaces, being formed by several trabeculas that give a porous aspect to the tissue. The spongy bone is the one with minor weight and has grind shape, with bone spaces where it is found the

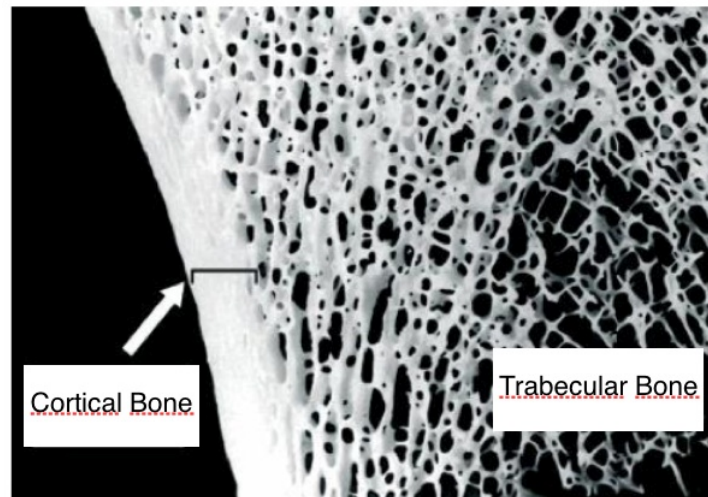


Figure 2: Trabecular and Cortical Bone (Cowin, 2001)

medullary bone. Generally, this kind of bone tissue is localized in the extremities of the long bones where is required high distributed resistance as shown on Figure 2.

The compact bone tissue has almost none medullar space, but a conjunct of canals where nerves and blood vessels go through. Inside the cylindrical cavity of the long bones is found the medullary bone. Generally, the cortical bone is found in regions of very high stresses. In long bones it is possible to identify three distinct regions: the epiphyses that corresponds to the extremities of the bones and is composed by trabecular bone tissue; the diaphysis which is located between the extremities and is composed, primordially, by cortical bone; and the metaphysis, located between the epiphysis and the diaphysis. The thickness of the layer of bone tissue and the resistance to rupture increase progressively going from the metaphysis to the diaphysis Neto *et al.* (1999).

In biomaterials area, the knowledge of the mechanical properties of the bone is extremely important, specially, in the development of prosthesis more adequate to human conditions. The bone is considered a material with great mechanical resistance at the same time that shows elasticity and lightness. it is a dynamic structure, alive and continually renewed through the life of the person, according to the type and intensity of the load. The cortical bone is frequently considered an orthotropic material. Orthotropic materials are a class of anisotropic materials characterized by three Young modulus according to the direction of the load, three shear modulus and six Poisson coefficients.

Besides the anisotropy, another important characteristic of the bone is the viscoelasticity. The viscoelasticity is the property of the materials that have both characteristics, viscous and elastic, when under deformation. Viscous materials resist to shear rate and stress linearly with time when a stress is applied. Elastic materials deform instantly when under tension and compression and return equally fast to the original state when the load is removed. Viscoelastic materials have elements of both properties. As such, exhibit stresses dependent of the strain and the strain rate.

Holanda *et al.* (1999) indicated that the collagen is the bigger responsible for the bone viscoelastic behavior. Contrary to the elastic materials, viscoelastic materials lose energy in a cycle of loading by thermal dissipation. Additionally, they are characterized for the capacity of having viscous strain under constant loads and relaxing under constant displacements. The different structures of the cortical and trabecular bone result in different mechanical properties that are highly variable with the species, age, anatomy, liquid content, among other factors.

The high anisotropic nature of the bone combined with its viscoelasticity and dynamic change of properties with time make the value of the mechanical properties vary from bone to bone. Even thought, Table 1 presents an average value of mechanical proprieties of human bone.

Table 1: Mechanical Proprieties of Human Bone

	Cortical Bone	Trabecular Bone
Limit Stress to Tension Resistance	150 MPa	3 MPa
Limit Stress to Torsion Resistance	49 - 68 MPa	15 MPa
Shear Modulus	2840 - 4040 MPa	8 - 40 MPa
Poisson Coefficient	0,2 - 0,5	0,01 - 0,35

Machining bones is not a usual topic of research because medical procedures are quite simple and well known during orthopedic surgery. When it concerns manufacturing of threads made of bones to be used later on surgery, the situation changes and there is a need of knowledge in order to make a proper workpiece.

3. THREAD MANUFACTURING

Thread manufacturing is the operation in charge to make internal or external threads, with normalized shapes and dimensions, on cylindrical or revolved conical surfaces. Threads can be produced by foundry, injection, mechanical forming and machining. When produced by forming, the thread has a higher resistance because of the strain hardening. Additionally, the absence of chip formation and long life of the tool are other advantages of the forming process (Bretas *et al.*, 2009). In opposite, the precision achieved by this technique is not capable of outdoing the one achieved by machining operations. There are a lot of different machining processes that can be used to manufacture the thread: turning, taps and dies machining, thread whirling, grinding and milling. The choice of the proper process takes into consideration several factors, such as: dimensional tolerance of the thread, geometry of the part and, mainly, the diameter of the thread (Bretas *et al.*, 2009).

3.1 Thread Milling

Thread milling has becoming very popular in the last years as an alternative to other forms of manufacturing thread. Threads manufactured by this process can be done with single cutting edge or multiple cutting edges as in thread milling (Araujo *et al.*, 2005a).

Thread milling tools are similar to taps, but its operation is completely different. Contrary to tap, thread milling tools do not have chamfer and are capable to manufacture the whole length of the thread, even in materials that were hardened, without sacrificing its profile, finishing and precision. Moreover, the feed per tooth speed of the tool does not depend of the thread pitch (Smith, 2008). This method requires a CNC machine able to perform helical interpolations. In comparison to the conventional thread manufacturing processes, thread milling has the following advantages: provides good surface finish and precision; generates smaller chips; allows tolerance adjustments through exact calculations; allows full thread until the bottom of the hole; provides prolonged tool life; it is adequate for most materials; the same milling tool can be used to different diameters, as long as the pitch is the same; the same tool can be used to both left and right thread and both internal and external thread; and allows dry cutting.

The thread manufacturing in holes with thread milling tools can be divided in five stages. The first stage consists in the positioning of the tool from the start position to the central axis of the hole. Then, the tool is dislocated to the position where will start the cutting. Once in place and rotating at the selected spindle speed, the tool moves axially with a determined feed velocity until an appropriate cutting radial depth. In the third stage, the tool executes an up helical movement in the hole to a complete revolution of the tool axis. The number of threads made in one revolution of tool axis matches the number of threads initially engaged in the wall of the part. If the creation of threads to the total depth of the hole is not achieved in one round, the stages 1 to 3 must be repeated until all the threads are completed. Once the thread is manufactured, the tool moves back to the center (fourth stage) and out of the hole (fifth stage) Araujo *et al.* (2005b).

3.2 Thread Milling Penetration Strategies

There are different methods of milling tools approximation to start the thread cutting, named thread penetration strategies (Fromentin and Poulachon, 2010). Some CNC machines can be equipped with the specific cycles for this operation, however, machines with simpler commands demand different programming strategies.

The simple form of radial positioning of the tool is the linear entrance. According to Smith (2008), this strategy is characterized for generating a high contact angle between the material and the peripheral diameter of the tool. As a result, there is a high load situation on the tool which, depending on the material, can lead to long chips. Besides, the linear entrance is not advisable to precision threads because it leaves marks on the entrance point of the tool and it produces interferences and over machining of parts.

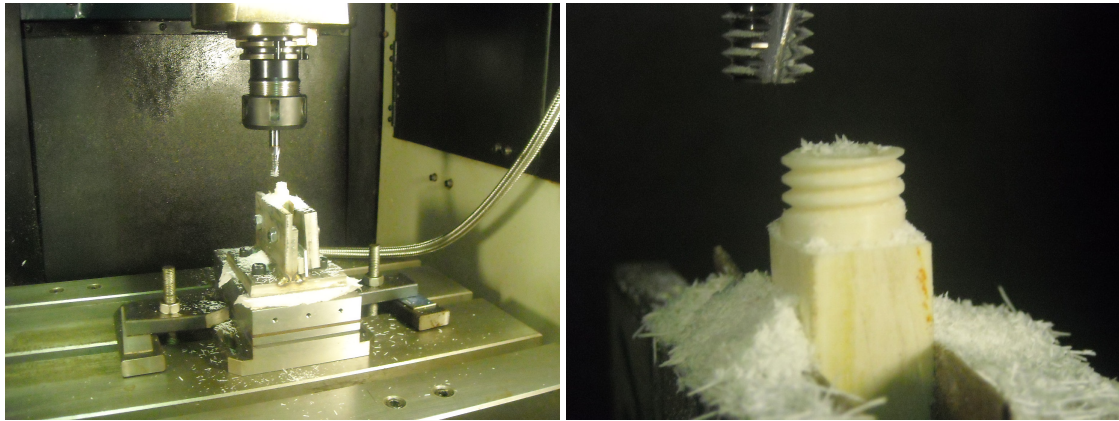
Most used penetration strategies are half penetration and quarter circle penetration. On those cases the tool axis moves with circular trajectory to the initial point of the full circle revolution that will machine the complete thread. A full study of interference on thread penetration was developed by Vishal *et al.* (2010).

4. EXPERIMENTAL PROCEDURE

At this work it was verified the viability of manufacturing a thread on a piece of bone removed from a bovine femur. The experiment (Figure 3(a)) consisted of the manufacturing of the thread in a CNC machining center, using a thread milling tool. Moreover, were performed the measurement of the cutting forces involved in the process.

The knowledge of the cutting forces and the studying of their components are of great importance to the evaluation of the potency required to execute the cutting. Besides, it is an important criterion for the determination of the machinability of a given material.

In the specific case treated at this work, it is important to observe the behavior of the specific cutting pressure when cutting a material as an insert of bovine bone. The forces which act on the cutting edge strongly depend on the condition of the chip-tool interface. The principal factors which influence the cutting forces are: the workpiece material, the material and the geometry of the tool, cutting area, cutting speed, chisel angle and cutting fluid condition.



(a) Experimental Set-up

(b) Machined Workpiece

Figure 3: Experiments for measuring the cutting forces (Souza, 2010)

4.1 Experimental Stages

The experiment is divided in five steps, as described:

- I** - Preparing the Workpiece: Extraction of a piece of material from the bovine femur to be milled, cleaning and cutting of the bone.
- II** - Fixation of the sample: Manufacturing of a fixture device that allowed the attachment of the bone insert to the dynamometer.
- III** - Circular milling.
- IV** - Thread milling with the measurement of the cutting forces.

4.1.1 Stage I - Preparing the Workpiece

Before the start of the milling process, the bone had to pass through some preliminary procedures of cleaning and whitening. First, the bone was cut to remove the epiphysis (cutting the longitudinal edges) that is mostly made of trabecular bone tissue. Next, it was done the removal of the medulla that stays inside of it in a place called medullar cavity. After these two stages, the bone was boiled in water and bar soap for about two hours. This procedure facilitates the removal of all the residues which still are stuck on the surface. Then, the bone passed through alternate baths of sodium hypochlorite (bleach), water, hydrogen peroxide (oxygenated water) and water again. Last, the bone was exposed to sunlight for a couple of days. After this procedure, it got properly cleaned and bleached.

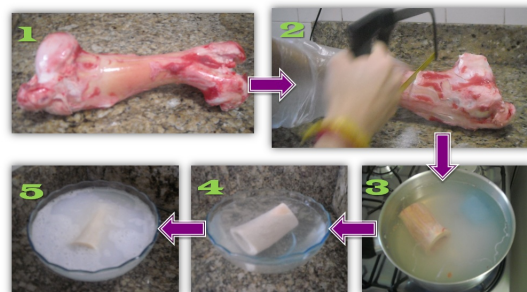


Figure 4: Preliminary Bone Preparation (Souza, 2010)

In Figure 4, there is the illustration of each of the stages made on the preparation of the bone to be milled: (1) bone in its original state; (2) removal of the extremities; (3) boiling in water and soap; (4) bath in sodium hypochlorite solution and (5) bath in hydrogen peroxide.

The final shape of the sample was obtained using a hacksaw with a Bi-Metal blade. Despite the extremely irregular shape of the bone, it was tried to obtain a shape close to a parallelepiped with a base of 10 mm X 10 mm and height of 35 mm.

An observation can be made concerning possible alterations that the cleaning procedure and bone preparation may cause in its mechanical property. Previous works about bone machining used ungreased or freeze-died bone tissue. Freeze dying or cryodesiccation is a process of dehydration used to preservation where the material in question is frozen and the water is removed by sublimation, that is, not passing for the liquid phase. According to Bento (2003), in terms of mechanical behavior the cortical bone is more affected by the freeze-dying than the trabecular bone.

Cowin (2001) cites experiments that confirm a reduction of the elasticity modulus, of the deformation and of the rupture energy, in cortical bone freeze-died at -30°C and, later rehydrated in salty solution for two hours at temperature of 20°C . Furthermore, according to Bento (2003), a sample of bone, when removed from a living organism will inevitably suffer alteration of its mechanical properties. With the death of the cells of the bone tissue, will be observed temperature variation and, principally, humidity alteration. If the bone is kept in a humidity and temperature state compatible with the one of the body, will be produced enzymes that can deteriorate the organic and mineral matrix of the bone. An alternative that would avoid this problem would be the freezing of the bone. At last, it is worth pointing that the method of cleaning of the material used in the present work is very simple, but it satisfies the objective of the experiment which is the studying of the machining of an insert of a bovine bone. In the case of the manufacturing of implants, more sophisticated techniques would be necessities, like, for an example, the irradiation of gamma rays to eliminate infectious organisms.

4.1.2 Stage II - Fixture Design

The fixture manufactured in the stage B was made from an inox steel 304 plate with dimensions of 300 mm X 100 mm and thickness of 1/2 in. The base of the fixture had four holes which allowed attach it to the dynamometer using four Allen M8 screws. The efficient fixation of the bone was harder because of its very irregular shape. The fixture was composed of a base and two vertical plaques, between which had a space reserved for the positioning of the bone. The fixation of the bone insert was given by the squeezing of one plaque against the other through two Allen M8 screws.

4.1.3 Stage III - Circular End Milling

Machining of a cylinder from piece of material extracted from the bone Before the performance of the threading, it was necessary to mill the material in order to obtain a cylinder with the diameter equal to the external diameter of the thread. The process was done in the ROMI Polaris V400 CNC machining center, using as tool an end milling tool. It was not applied any working fluid during the process. Basically, the tool performed a circular interpolation in the XY plane in counterclockwise direction. The machining happened in several passes. To each pass, the program was executed once making the correction of the radius.

In order to manufacture a cylinder with 10 mm of diameter, it was used an end milling tool with 10 mm of diameter, (INDAÇO model 1501). The geometric characteristics of the tool are showed in Table 2.

Table 2: Parameters of the end milling process

Milling tool diameter	Total length	Number of tooth	Helix angle
10 mm	22 mm	4	30°
Spindle speed	Depth of cut	Feed speed	Working fluid
240 RPM	7 mm	16 mm/min	Dry Cutting

4.1.4 Stage IV - Thread Milling

On this stage, the Mmanufacturing of a right handed type thread on the workpiece is done by the milling process using a thread milling tool. The operation was performed by the same machine tool of the previous stage. Again, it was not used any working fluid. Also at this stage, it was measured cutting forces with a piezoelectric dynamometer.

As the machine tool used to execute the experiment did not have helical interpolation, the operator used small circular counterclockwise interpolations on the XY plane, alternated with linear displacements on the Z axis upwards. At each 120° arch made by the tool, it went up vertically 0.05 mm. With that, it was possible to make the tool to perform an approximated helical trajectory.

A radial penetration method with quarter circle was chosen. Such technique avoids that marks are left on the entrance point of the tool. The assembled experimental set-up is showed in Figure 3(a) and 3(b). For the manufacturing of the thread was used a Vargus M10 X 1.5 mm thread milling tool and the parameters shown on Table 3. For the selection of the cutting parameters (cutting speed and feed), Sandoval *et al.* (1996) was taken as reference. The experiment executed by Sandoval *et al.* (1996) was the one the used the smaller cutting speed comparing to the other references. The experiment showed that there the operation can be done without working fluid in this range.

For the measurement of the forces involved in the manufacturing of the thread was used a Kistler dynamometer, model 9257 AB, connected to a signal conditioner and amplifier, also from Kistler, model 5233A1. This set gives the cutting force components in X, Y and Z directions. A digital oscilloscope helped on the definition of the output value and on

Table 3: Parameters of the Thread milling process

Milling tool diameter	Thread	Number of tooth	Helix angle
8.2 mm	M10 X 1.5 mm x 15.0 mm	3	30°
Spindle speed	Feed per tooth	Cutting speed	Working fluid
240 RPM	0.02 mm	7.5 m/min	Dry Cutting

the behavior of the signal of the three components before the acquisition. The acquisition of the signals was performed connecting the signal conditioner to a data acquisition board. This board transmitted the signals of the force components to a microcomputer using the software LabVIEW with acquisition rate of 1 KHz.

5. EXPERIMENTAL RESULTS

5.1 Experimental Forces

During the manufacturing experiment of the thread, it was measured the cutting forces involved in the process. The dynamometer acquire forces F_X , F_Y and F_Z in table referential R_0 . R_1 , the cylindrical surface referential, and R_2 , which follows the spindle movement, are presented in Figure 5a.

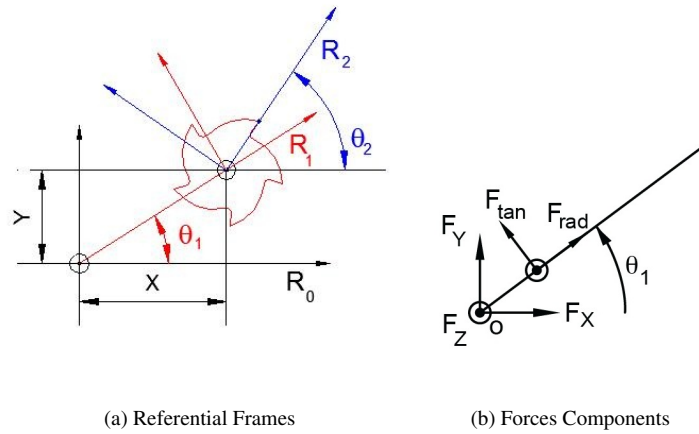


Figure 5: Forces decomposed in three Referential Frames: R_0 , R_1 and R_2 .

In order to transfer to the machined surface referential R_1 , a rotation matrix is applied as a function of θ_1 , the position angle related to the tool axis position and the table referential, as shown in Figure 5a. Note that θ_1 is a function of tool positions X and Y .

$$\vec{F}_r = \begin{bmatrix} F_{rad} \\ F_{tan} \\ F_Z \end{bmatrix} = \begin{bmatrix} \cos(2\pi - \theta_1) & -\sin(2\pi - \theta_1) & 0 \\ \sin(2\pi - \theta_1) & \cos(2\pi - \theta_1) & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} F_X \\ F_Y \\ F_Z \end{bmatrix} \quad (1)$$

All signal was taken during complete machining process but only steady state and full engagement during machining is considered in all data experiments, using the θ_1 range from $180^\circ < \theta_1 < 340^\circ$. The steady state range is displaced in Figure 6a.

It is possible to verified force spikes (negative) in the Z direction that coincide with a fall of the modulus of the forces in X and Y. This behavior is the consequence of the machining procedure adopted in the experiment: alternated linear interpolations with linear movements in Z. In order to be precise and disregard the linear movements in Z, thread milling data was extracted. Although the peaks in Z were important to extract θ_1 calculation from CNC code. Forces in one revolution in R_0 are presented in Figure 6b.

Using θ_1 angles, force components in R_1 were calculated using Eq. 1 and it is displayed in Figure 7 in different tool positions.

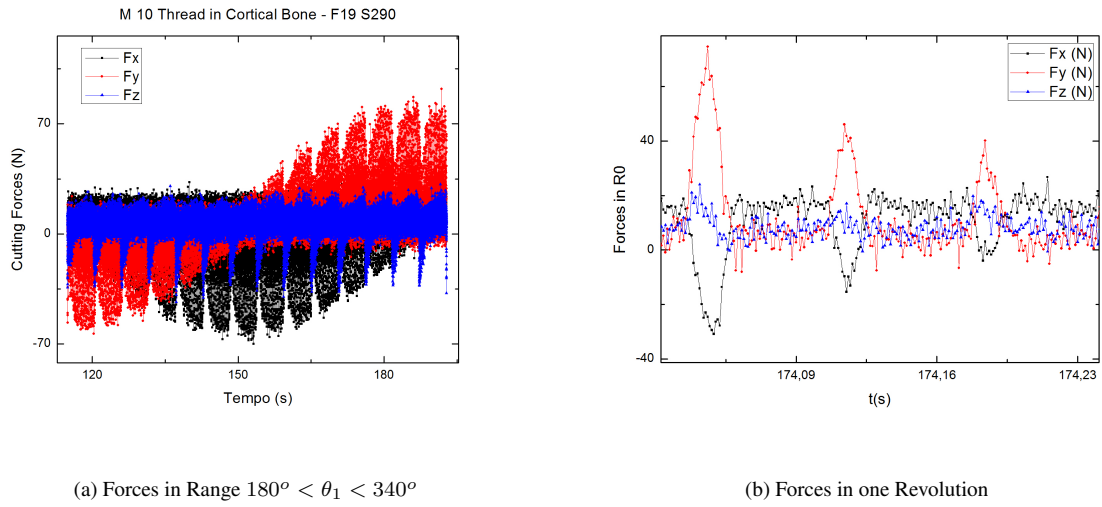


Figure 6: Experimental Thread Milling Forces in R_0

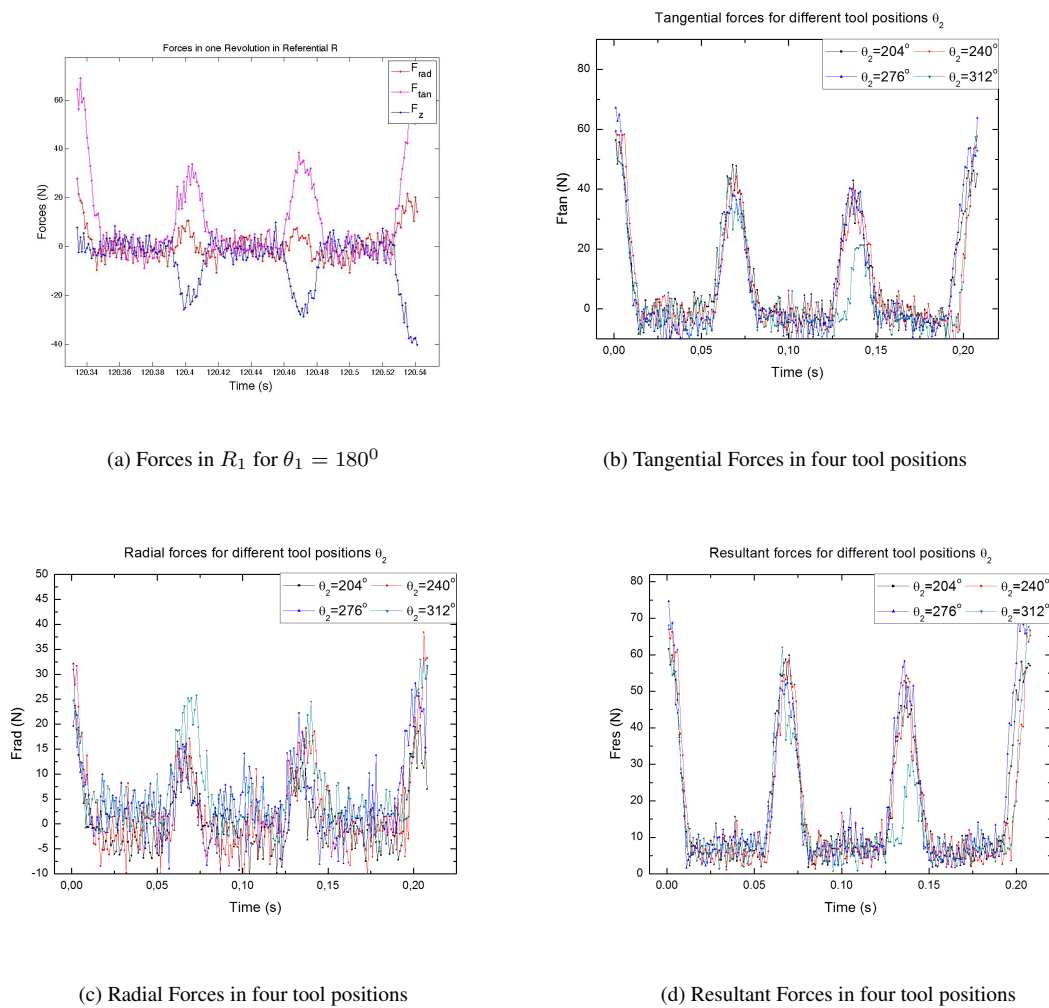


Figure 7: Forces in One Revolution Forces for different θ_1

In the range ($180^\circ < \theta_1 < 340^\circ$), 14 revolutions were taken for the statistical analysis considering the three maximums related to each thread flute. Those revolutions were equally spaced to guarantee independence of axial depth of cut a_{doc} variation. The average of all maximum resultant forces for each flute \bar{F}_{res} is presented in Table 4. Using the R_1 componentes \bar{F}_{rad} and \bar{F}_{tan} were calculated.

It is also computed values of forces in those different θ_1 positions divided by the local a_{doc} and the average of these relations are presented in Table 4 as \bar{F}_{res}/a_{doc} , \bar{F}_{rad}/a_{doc} and \bar{F}_{tan}/a_{doc} .

Table 4: Results for average maximum of data from 14 θ_1 locations

\bar{F}_{res}	\bar{F}_{res}/a_{doc}	$\frac{\bar{F}_{res}}{a_{doc}} \cdot a_{doc}$
56.8 ± 5.4 N	11.43 ± 1.03 N/mm	56.8 ± 5.1 N
\bar{F}_{tan}	\bar{F}_{tan}/a_{doc}	$\frac{\bar{F}_{tan}}{a_{doc}} \cdot a_{doc}$
46.9 ± 4.2 N	9.42 ± 0.74 N/mm	46.8 ± 3.7 N
\bar{F}_{rad}	\bar{F}_{rad}/a_{doc}	$\frac{\bar{F}_{rad}}{a_{doc}} \cdot a_{doc}$
23.9 ± 4.7 N	4.84 ± 1.08 N/mm	24.11 ± 5.4 N

5.2 Thread Quality

There are not enough studies today to determine the ideal roughness that a bone implant must have. However, it is known that this roughness must insomuch that it favors the bone integration process. According to Mora (2000), a rough surface, probably, facilitates the structural and functional connection between the alive bone and the surface of the implant. A picture of the profile of a manufactured thread can be viewed in the Figure 8. The image was obtained amplifying the thread on a microscope. From this profile, it was calculated the average roughness Ra. The result was $43 \mu\text{m}$. With the help of a caliper with 0.05 mm resolution, was measured the external diameter of the machined thread. The value that was verified was 9.90 mm.



Figure 8: Microscopic Picture of one Manufactured Thread (Souza, 2010)

6. CONCLUSIONS

The experiment presented that is possible to manufacture a thread in a piece of bovine cortical bone using as threading technique the thread milling with a tool with multiple cutting edges. Through the experiment, was noted that the behavior of the bovine bone during the machining does not differ much from the one showed by fragile materials, with the formation of short loose chips. It must be emphasized that the piece of bone selected to the operation was extracted from the cortical (compact) bone, with no vestiges of trabecular bone tissue which has a porous structure, that could result in a damaged part.

It is especially important to control the cutting temperature when it is machining bones of live organisms, so that there is not the necrosis of the bone tissue and cells. The cellular necrosis of the bone starts at a temperature of 50°C and irreparable damages occur at temperatures bigger than 70°C . Thermal effects lead to the denaturing of the enzymes and proteins of the cellular membrane, resulting on the dead of the cell. Besides, the thermal damages to the bone can be attributed to both the magnitude of the temperature and to the time of exposure to high temperature (Sugita *et al.*, 2005). To obtain lower cutting temperatures and, consequently, smaller possibilities of thermal damages, it must be selected cutting speed as smaller as possible and use salty solutions as working fluid. For the applied cutting conditions, the bone showed to be a material of easy machining and it was not verified as excessive heating that could damage the material. Sugita *et al.* (2005), in their study on cutting temperature of bone machining, verified a very small influence of the cutting condition, as speed cutting and feed per tooth, in the temperature. Also, it was verified that raising substantially the cutting speed, the variation of the cutting forces was very small (Sugita *et al.*, 2009b,a). It is important to point out that in the

case of the present work, the cutting temperature and the possible necrosis of the tissue are not critical aspects, once the material machined is a piece of dead bone.

The bone machined showed good machinability and, for the cutting parameters selected, it was not verified over heating that could damage the material or the tool. Also, it was verified that is possible to manufacture a normalized thread in a piece of bovine cortical bone. In the performed experiment, it was produced a metric thread ISO M10 with pitch of 1.5 mm. The cutting forces involved in the process were measured using a dynamometer and the maximum forces in each flute is analyzed and presented in the article. Bending component force in the tool showed to be very low comparing to the resultant force. Forces Tangential to the thread surface have values around 50 N while resultant force achieved 57N. The force profile is the analogous to other metallic materials.

Regarding the quality of the machined surface, the result obtained is closely linked to the method used to get the tool to perform the trajectory. As claimed, the machining center is not prepared to perform helical interpolations and alternating small circular interpolations with small upwards movements was used. This alternate movement introduced on the part little marks that would not exists if it was used helical interpolation.

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9. DIREITOS AUTORAIS

Os autores são os únicos responsáveis pelo conteúdo do material impresso incluído no seu trabalho.