SIMULATING THE BONE-TITANIUM INTERFACIAL CHANGES AROUND TRANSFEMORAL OSSEOINTEGRATED IMPLANTS USING PHYSICAL MODELS AND MODAL ANALYSIS

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ABSTRACT

Non-invasive vibration analysis is being considered as a method to monitor the healing progression of femoral implants in transfemoral amputees. Studies to date have successfully detected gross alterations in the physical properties of the interface region of physical bone-implant models using vibration techniques. This paper describes the development of a series of physical models which simulate the incremental bone-implant interfacial changes during progressive osseointegration. The capability of modal analysis to detect the changing interface conditions is investigated. The model resonant frequencies and their mode shapes altered due to the different interface conditions. Higher resonances were shown to be more sensitive to interface changes than the fundamental frequency. The findings demonstrate the potential of modal analysis for this application and the technique warrants further investigation.

Keywords: osseointegration, modal analysis, transfemoral amputation, physical model

1. INTRODUCTION

An alternative to using a prosthetic socket for above knee (transfemoral) amputees is transfemoral osseointegration (TFOI) (Hagberg and Brånemark 2009). A titanium implant is inserted into the medullary canal of the amputated femur; the implant protrudes through the skin and connects directly to the prosthetic limb removing the need for a socket (Ward and Robinson 2005). Studies have reported several advantages of TFOI compared to using a socket (Hagberg 2005; Hagberg, Brånemark et al. 2008; Sullivan, Uden et al. 2003) and it can be particularly appropriate for amputees that experience skin problems due to socket wear, for those with a short residual limb and those with an active lifestyle (Hagberg, Brånemark et al. 2008).

However, it can take twelve to eighteen months for the implant to integrate with the bone and for an amputee to be able to load bear and therefore be fully rehabilitated (Ward and Robinson 2005). The long rehabilitation time is a significant disadvantage of TFOI and may be impeding the wider adoption of the technique.

Vibration analysis techniques are being investigated as non-invasive methods of assessing the degree of bone-implant integration (known as osseointegration (OI)) (Cairns, Adam et al. 2011; Shao, Xu et al. 2007; Swider, Guérin et al. 2009; Xu, Shao et al. 2005). Physical models have been used to simulate different interface conditions between the femur and implant that may occur during osseointegration. The different physical properties at the femur-implant interface can be identified by measuring the changes in the dynamic properties of the system (Cairns, Adam et al. 2011; Shao, Xu et al. 2007; Xu, Shao et al. 2005). The longer-term aim of the vibration analysis research is to determine when the implant is able to withstand physiological load by monitoring the dynamic properties of osseointegration progression and potentially reduce the overall rehabilitation time.

Despite the reported success of the vibration techniques, the physical models developed in previous studies (Cairns, Adam et al. 2011; Shao, Xu et al. 2007; Xu, Shao et al. 2005) compare interface conditions which represent gross changes at the femur-implant interface *in vivo*. The interface conditions developed in the author's previous work (Cairns, Adam et al. 2011) were intended to "represent extremes of the spectrum of implant integration with the bone" and were used to establish the feasibility of the modal analysis methodology employed. Further work is required to develop more appropriate interface condition models and ascertain if modal analysis is capable of detecting femur-implant interfacial changes that are more representative of the *in vivo* scenario.

This research details a series of composite femur-implant physical models of TFOI developed to represent the known histological and mechanical properties at the femur-implant interface during osseointegration. Modal analysis is then conducted using the models to establish if the technique remains

capable of detecting changes at the interface between the implant and the femur.

2. METHODOLOGY

2.1. Physical Model Development

It is important to develop physical models that represent the real structure as realistically as possible. To this end, literature on the physiology of the bone-titanium interface was reviewed to identify key stages of transfemoral osseointegration. Physical models were then developed to simulate the mechanical characteristics of the femur-implant interface associated with the key stages.

The femur model and implant model were common to all physical models, only the interface condition was altered to represent the different stages of OI. Fourth generation large composite femurs (Sawbones model 3406, Pacific Research Laboratories Inc, WA, USA) were used as the femur model. The femurs were cut to a length of 237mm (distally from the femoral head), replicating the amputated femur (Figure 1(a)). Implants were machined from commercially pure titanium rod with a threaded section 80mm long, 19mm outer diameter and 1.75mm male thread pitch. The profile of the implants changed to a cylindrical section 60mm long, 15mm outer diameter (the dimensions used clinically). Flats were machined on the cylindrical section and a 2.5mm threaded hole was machined in one flat (Figure 1(b)) to allow attachment of the excitation hardware (detailed in section 2.2).

A physiological model found in the literature outlines four stages of bone-titanium osseointegration (Brånemark, Gröndahl et al. 2005). The Brånemark physiological model, other literature supporting the model (including histological and mechanical characteristics of the interface) and the physical model interfaces developed to represent the four stages are detailed in sections 2.1.1 to 2.1.4.

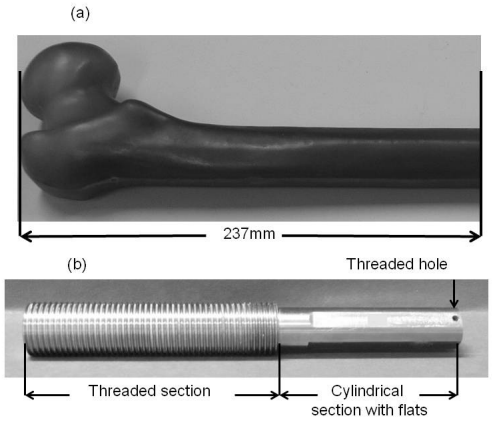


Figure 1: (a) the cut composite femur and (b) the titanium implant

2.1.1. Stage 1 Physical Model

In stage 1 of the Brånemark physiological model, immediately after the implant is inserted, there are areas of bone-implant contact and also gaps filled with haematoma. The bone surrounding the implant is damaged while the bone further away from the implant is healthy. A combination of bone-implant contact and haematoma filled gaps has also been reported in numerous long bone animal studies immediately after implant insertion (Dhert, Thomsen et al. 1998; Franchi, Fini et al. 2005; Linder, Albrektsson et al. 1983; Sennerby, Thomsen et al. 1993; Uhthoff 1973).

To replicate the combination of bone-implant contact and gaps, the medullary canal of the femur was pre-threaded using a CNC machine to achieve an implant insertion torque of 5Nm. The insertion torque value 'at first implant insertion' was chosen to be lower than 12Nm; the torque the implant is tightened to six months after insertion (Ward and Robinson 2005).

Mechanical testing of the bone-implant interface immediately after implant insertion has demonstrated negligible tensile strength (Kitsugi, Nakamura et al. 1996; Steinemann, Eulenberger et al. 1985) and shear strength (Brånemark, Ohrnell et al. 1997; Brånemark, Ohrnell et al. 1998; Ivanoff, Sennerby et al. 1996; Johansson 1987; Rubo de Rezende and Johansson 1993; Sennerby, Thomsen et al. 1993). Therefore no additional interface materials were used to bond the implant to the femur.

To assemble the Stage 1 model the implant was inserted in the threaded canal of the femur using a dial torque wrench which continuously measured the 5Nm insertion torque. The implant was inserted to a depth of 90mm measured from the cut end of the femur. The interface simulation is summarised in Table 1 and the model is illustrated in Figure 2.

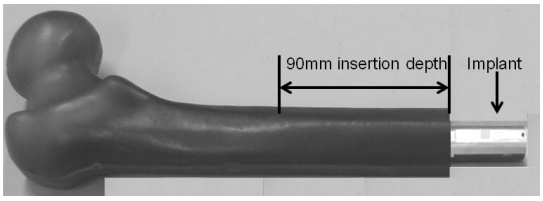


Figure 2: Assembled Stage 1 physical model showing implant inserted in pre-threaded femur

2.1.2. Stage 2 Physical Model

At stage 2 of the physiological model complete contact between the implant and the bone is achieved: the haematoma forms new bone to fill the gaps and the surrounding damaged bone heals. This process of new bone formation and resorption/replacement of damaged bone is supported by the findings of numerous long bone animal studies (Buma, van Loon et al. 1997; Dhert, Thomsen et al. 1998; Franchi, Fini et al. 2005; Linder, Albrektsson et al. 1983; Sennerby, Thomsen et al. 1993; Uhthoff 1973; Ysander, Brånemark et al. 2001). This process is estimated to take over four months in the human (Roberts, Turley et al. 1987). It is thought that modal analysis would need to be capable of detecting interfacial changes throughout this period in order to be a useful technique. Therefore two physical models were manufactured to represent different time

points in the formation of mature bone; Stage 2-intermediate (Stage 2-int) and Stage 2-end.

The tensile strength of the interface remains minimal after bone remodelling (Kitsugi, Nakamura et al. 1996; Steinemann, Eulenberger et al. 1985) while the implant removal torque increases (Brånemark, Ohrnell et al. 1997; Brånemark, Ohrnell et al. 1998; Ivanoff, Sennerby et al. 1996; Johansson 1987; Rubo de Rezende and Johansson 1993; Sennerby, Thomsen et al. 1993).

To represent an intermediate point with immature bone in complete contact with the implant, an additional material was inserted at the interface of the Stage 2-intermediate model. The interface material was a liquid to solid resin (Palapress, Heraeus Kulzer GmbH, Germany) with a lower elastic modulus than the femur (16GPa for composite femur; 2.1GPa for resin). The size of the necrotic bone region undergoing remodelling ranges from 0.5mm to 40% of the bone radius (Albrektsson 1985; Buma, van Loon et al. 1997; Roberts, Turley et al. 1987). Therefore the femur was bored out to a diameter 3mm larger than the implant diameter.

To assemble the Stage 2-intermediate model the resin was poured in to the femur canal and then the implant was inserted to a depth of 90mm so that the resin filled the gap around the implant. When cured the resin did not adhere to the implant therefore complete femur-implant contact was achieved with negligible interface tensile strength.

To represent an end point with mature bone in complete contact with the implant, the medullary canal of the femur was pre-threaded using a CNC machine to achieve an implant insertion torque of 20Nm. This value of insertion torque was chosen to be larger than 12Nm required to attach components to the implant (Ward and Robinson 2005). No additional materials were used therefore the interface had negligible tensile strength.

To assemble the Stage 2-end model the implant was inserted in the threaded canal of the femur to a depth of 90mm using a dial torque wrench which continuously measured the 20Nm insertion torque. The interface simulations are summarised in Table 1.

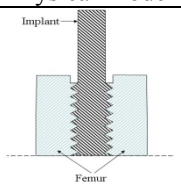
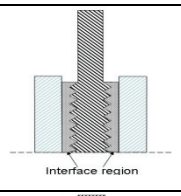
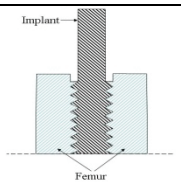
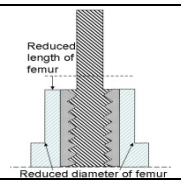
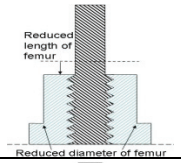
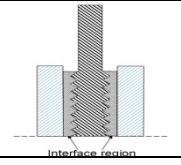
2.1.3. Stage 3 Physical Model

At stage 3 of the physiological model, the healthy revascularised bone can now withstand load and remodels due to the loading stimulus applied. Animal studies that have applied loading protocols to implants (after an initial unloaded healing period) provide further evidence that bone remodelling occurs due to load stimulus (Brunski, Hipp et al. 1989; Duyck, Ronold et al. 2001; Hoshaw, Brunski et al. 1994). Furthermore there is some clinical evidence of bone surface remodelling around the TFOI implant *in vivo* when load is applied to the implant (Xu, Shao et al. 2005). To the author's knowledge there is no information on the tensile and shear strength of the bone-titanium interface after implant loading.

As the process of bone remodelling due to load stimulus would be similar to remodelling of damaged bone in Stage 2, no additional interface models were made to replicate bone remodelling. However the surface modelling resulting in bone thickness variations was thought to be an important clinical observation. Therefore the Stage 2 physical models were modified to simulate external surface resorption and named Stage 3-intermediate (Stage 3-int) and Stage 3-end

To represent the distal bone resorption (Xu, Shao et al. 2005) the femur length was reduced by 10mm on both models. To represent the cortical wall tapering along half the implant length the femur diameter was reduced to 26mm over 40mm length. The interface simulations are summarised in Table 1.

Table 1: Summary of the Mechanical Characteristic of each Physical Model and a Schematic of the Model

Stage	Mechanical Characteristics of Physical Model	Schematic of Physical Model
1	Femur pre-threaded to produce implant insertion torque of 5Nm. No adhesion between implant and femur. Low shear strength at interface	
2-int	Femur bored out to radius 1.5mm larger than implant. Gap filled with resin. No adhesion between resin and implant. Increase in implant removal torque	
2-end	Femur pre-threaded to produce implant insertion torque of 20Nm. No adhesion between implant and femur. Increase in implant removal torque	
3-int	Stage-2-intermediate modified. Femur length reduced by 10mm. Femur diameter reduced over 40mm implant length	
3-end	Stage-2-end modified. Femur length reduced by 10mm. Femur diameter reduced over 40mm implant length	
4	Femur bored out to radius 1.5mm larger than implant. Gap filled with silicone. No adhesion between silicone and implant	

2.1.4. Stage 4 Physical Model

Stage 4 of the Brånemark physiological model refers to the formation of non-mineralised connective tissue between the bone and implant instead of healthy bone; this can occur in unsuccessful cases. Fibrous tissue

encapsulation around titanium implants placed in the femur of two animal studies verify the possibility of Stage 4 in the physiological model (Thomas and Cook 1985; Thomas, Kay et al. 1987). To the author's knowledge there is no information on the tensile and shear strength of the fibrous tissue-titanium interface that is known as unsuccessful osseointegration.

To simulate connective tissue encapsulation of the implant an additional material was inserted at the interface of the Stage 4 model. Fibrous tissue surrounding an implant has been mechanically tested (Hori and Lewis 1982) and a silicone elastomer found to have similar properties (Waide, Cristofolini et al. 2004). Therefore the same grade of liquid-to-solid silicone was used as the interface material (Sylgard 184, Dow corning Corporation, U.S; 2.6MPa elastic modulus). The thickness of the fibrous tissue region around an implant is greater than 1mm (Waide, Cristofolini et al. 2003). Therefore the femur was bored out to a diameter 3mm larger than the implant diameter (1.5mm thickness of silicone).

To assemble the Stage 4 model the silicone was poured in to the femur canal and then the implant was inserted to a depth of 90mm so that the silicone filled the gap around the implant. When cured the silicone did not adhere to the implant therefore the model had negligible interface tensile strength. The interface simulation is summarised in Table 1.

2.1.5. Femur Boundary Condition model

The boundary condition of the amputated femur *in vivo* is provided by the acetabulum and the connection of the muscles and soft tissue at the femoral head. This was simulated in the physical models by encapsulating the femoral head in a rectangular block of liquid-to-solid resin (Palapress, Heraeus Kulzer GmbH, Germany; 2.1GPa elastic modulus) and clamping the block to create a cantilever. A similar resin block has been used to simulate the boundary condition in the modal analysis of the fractured tibia (Nikiforidis, Bezerianos et al. 1990). Therefore the cantilevered boundary condition (Figure 3) was considered an acceptable first attempt at representing the *in vivo* boundary condition at the femoral head.

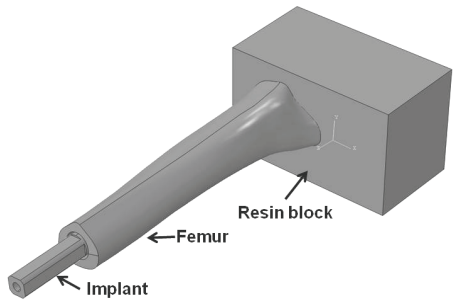


Figure 3: Resin block boundary condition encapsulating the femoral head

To create the resin block (dimensions 120x75x65mm), the femur was fixed in a custom-made mould which provided a minimum resin thickness of 10mm around the extremities of the femoral head. The resin was poured into the mould and allowed to cure. The physical model was then removed from the mould ready for modal analysis.

2.2. Modal Analysis

Forced excitation was applied to the models using an electromagnetic shaker driven by a power amplifier (part numbers 4810 and 2706 Bruel&Kjaer, Naerum, Denmark). The shaker methodology has been previously evaluated for this application using less complex models (Cairns, Adam et al. 2011). A signal generator (33120A, Agilent Technologies, CA, USA) input an excitation signal to the shaker.

The excitation was measured using a dynamic force transducer (0.028kg) and signal conditioner (part numbers 2311-500 and 4416B, Endevco, CA, USA). A Delrin stinger connected the shaker to the force transducer (Bruel&Kjaer, Naerum, Denmark). The other side of the force transducer was connected to the implant of the physical model using a screw connection in the threaded hole (Figure 1(b)).

The model response was measured using a single axis piezoelectric accelerometer (0.002kg) and a charge conditioning amplifier (part numbers 4393 and 2692-A-0S2, Bruel&Kjaer, Naerum, Denmark). The accelerometer was attached to the model using beeswax to allow the location to be easily changed.

The excitation and response signals were recorded using a 16-bit resolution data logger (USB-6259, National Instruments, NSW, Australia) connected to a personal computer (HP Intel ® Core™ 2Duo CPU 3.5GB RAM) using data acquisition software (LabVIEW SignalExpress version 2.5, National Instruments) and a sampling rate of 50kHz.

Figure 4 shows the coordinate system and excitation/response measurement sites identified along the model length. Seventeen response sites were identified for each femur-implant model. Site 17 was chosen as the excitation site.

The resin block was clamped to a steel base (dimensions 500x510x25mm) fixed to the laboratory floor. Sections of 12mm threaded rod were fitted through holes in the steel base and the resin block was fixed between the base and rectangular plates using nuts tightened to 16Nm. The experimental set up is shown in Figure 5. To maintain the correct alignment of the shaker and the model, the shaker was suspended on a spring over the physical model (Figure 5).

With the shaker attached to site 17 (dashed arrow in Figure 4; Figure 5) via the stinger and force transducer, the model was excited in the y-axis direction using a sine sweep signal (100Hz-5kHz frequency range, 500mV peak-to-peak amplitude and 5kHz per second sweep rate). The sine sweep parameters were optimised to obtain multiple resonant frequencies with adequate signal to noise ratio. The sweep was repeated

ten times and averaged in the data processing. The response was measured with the accelerometer attached to response site 1. The test was then repeated using the same excitation site but with the accelerometer attached to site 2-17 in turn. The y-axis testing was conducted on all six femur-implant models which are summarised in Table 1.

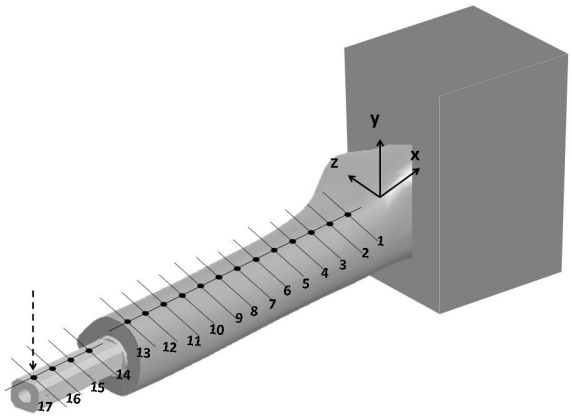


Figure 4: The coordinate system established for the models and the 17 response sites identified along the length of the model. Site 17 was chosen as the excitation site. The dashed arrow represents the y-axis excitation applied at site 17 used in all the modal testing.

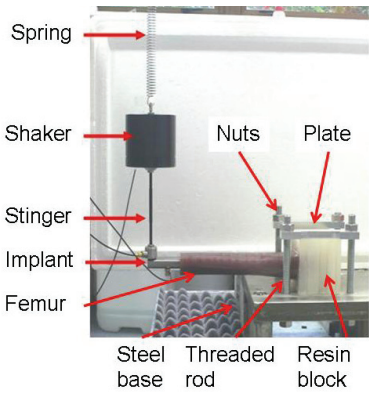


Figure 5: Experimental Set up of Physical Model Cantilevered to Steel Base. The shaker is suspended over the model using a spring and is connected to the implant at site 17 via a stinger and force transducer.

Customized analysis programs were written using MATLAB software (version 2007a, MathWorks Inc, Natick, MA, USA) to process the input and response signals and compute the frequency response function, accelerance, defined as the ratio of acceleration response to excitation force in the frequency domain. The MATLAB programs are detailed in Cairns (2010). Using a Fast Fourier Transform algorithm the accelerance function was computed from the tests performed at each excitation/response site combination (17 accelerance functions in total). These were used to calculate a mean accelerance function which was plotted against frequency to identify the resonant frequencies; resonances manifest as peaks on this type of plot. The peak picking method is illustrated in the author's previous work (Cairns, Adam et al. 2011).

Plots of the imaginary component of accelerance versus frequency at each excitation/response site combination were used to depict the mode shapes. The amplitude of the imaginary accelerance at a resonant frequency represents the displacement magnitude occurring at that site, while the sign of the amplitude indicates the positive or negative direction of the displacement (Avitabile 1999). Therefore by identifying the amplitude and the sign of the imaginary accelerance at each site (at a resonant frequency), the mode shape of the model can be determined. The resonant frequency values and mode shapes were compared between the physical models to ascertain if the different interface conditions could be detected.

3. RESULTS

3.1. Resonant Frequencies

Four resonant frequencies were identified for each physical model (Table 2). The signal to noise ratio was poor at frequencies over 3.5kHz and no resonances were identifiable above this frequency. It is likely that frequency changes relative to a baseline measurement recorded over time would be relevant *in vivo*. Therefore the percentage change in each resonant frequency from the baseline measurement (Stage 1 model) was calculated (Table 2). There is a maximum change of 5% and 7% in the fundamental and second frequency respectively due to the changing femur-implant interface properties. The third and fourth resonances are more sensitive to the alterations in interface condition; a maximum of 15% and 13% change in frequency respectively.

Table 2: Resonant Frequencies peak picked from Accelerance-Frequency plots for each physical model. Percentage Change in Resonant Frequency from the baseline Stage 1 model is shown in parenthesis [% change = ((Stage 1- Stage N)/Stage 1)*100, where N is physical model in table row]

	Resonant Frequency (Hz)			
Model (Stage)	1st mode	2nd mode	3rd mode	4th mode
1	221	662	988	2258
2-int	220 (0)	662(0)	1133(-15)	2435(-8)
2-end	223(-1)	669(-1)	1071(-8)	2394(-6)
3-int	230(-4)	671(-1)	1123(-14)	2450(-9)
3-end	231(-5)	669(-1)	1078(-9)	2541(-13)
4	214(3)	614(7)	986(0)	2240(-1)

3.2. Mode Shapes

The mode shape of the fundamental frequency of each physical model is illustrated in Figure 6. Typically mode shapes are normalised for comparison. The mode shapes have not been normalised so that the relative magnitude of the mode for each model can be compared (given that the excitation force was constant throughout

the modal testing). Data at site 13 is missing from the Stage 3 model because the femur length was reduced to simulate bone resorption. The fundamental mode shapes are similar to that of a classic cantilevered beam – approximately zero displacement at cantilevered end (site 1) increasing to maximum displacement at the free end (site 17). The modes have similar magnitude (imaginary accelerance value) as well as shape for all the physical models with the largest discrepancy evident in the Stage 2-int model.

The mode shape of the third frequency of each physical model is illustrated in Figure 7. It is evident that there are differences in the shape and magnitude of this mode across the models. The modes of the Stage 2 models are similar in shape and magnitude to the modes of the Stage 3 models. The Stage 1 model has a similar deformation pattern to Stage 2 and 3 at sites 1-8 and 14-17, but behaves differently at sites 9-13. The Stage 4 model is isolated on the plot with low magnitude displacement along the model length.

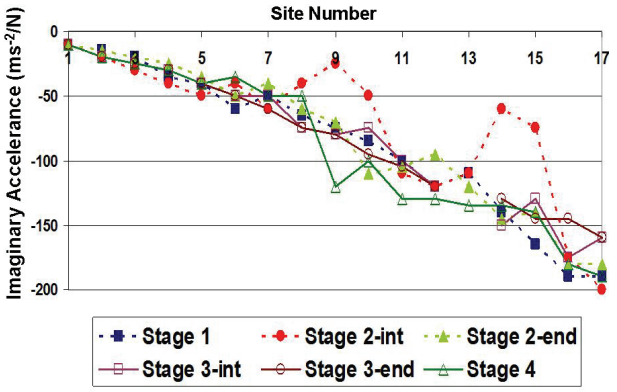


Figure 6: Fundamental Frequency Mode Shape of the Physical Models

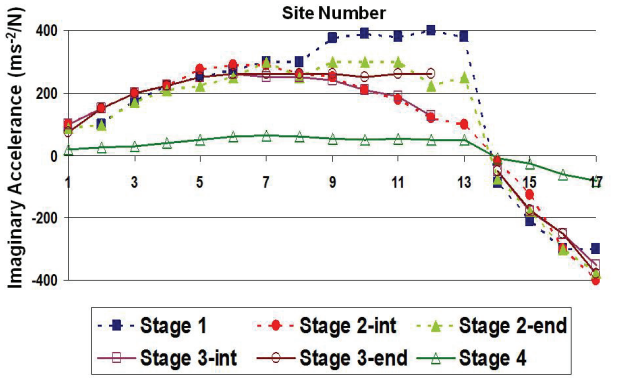


Figure 7: Third Frequency Mode Shape of the Physical Models

4. DISCUSSION

The first four resonant frequencies of the physical models changed due to the change in the physical properties at the femur-implant interface. In particular the third and fourth frequencies were more sensitive to the changing interfacial condition than the fundamental and second frequencies (greater percentage change in frequency; Table 2). This suggests that the higher frequencies may be more useful in the detection of osseointegration progression than the fundamental frequency. Therefore the modal analysis methodology reported here, which uses a broad frequency range excitation and detects multiple resonances, may offer greater functionality than impact excitation like that used in Shao, Xu et al. (2007) and Xu, Shao et al. (2005) where only the fundamental frequency of the femur-implant model was reported. This finding reinforces the author's previous work where sine sweep excitation delivered using an electromagnetic shaker was determined to be superior to impact excitation for this application (Cairns, Adam et al. 2011).

The small change in fundamental frequency (0-5%; Table 2) appears to contradict the author's previous work (Cairns, Adam et al. 2011) where the change in frequency between two femur-implant models with different interface conditions was 47% (estimated from the accelerance plot). The earlier investigation used different interface conditions and femur boundary conditions than those developed in the current study. Therefore, the results are not directly comparable. Nevertheless this finding suggests that the femur boundary condition has an effect on the magnitude of the frequency changes detected and this requires further investigation.

Arguably frequency changes compared to a baseline value would be useful for longitudinal measurements of osseointegration progression *in vivo*. In consideration of this, the changes in the third frequency are particularly interesting. The third frequency of the Stage 1, Stage 2-int and Stage 2-end models are quantifiably different. However the frequency change between the Stage 2 models and their Stage 3 counterparts (femur mass alterations simulating bone resorption but no interface change) is small (1%). This indicates that the third frequency may be capable of detecting different characteristics at the femur-implant interface but is not sensitive to bone mass changes. This finding may be important as the modal analysis technique *in vivo* would need to detect bone remodeling at the femur-implant interface due to load stimulus and not erroneously detect femur surface modeling which occurs at the same time. Furthermore there is no change in frequency when the Stage 4 model is compared to the baseline. This suggests that unsuccessful fibrous tissue formation could be distinguished from the progression of OI using the zero frequency change.

The fundamental frequency mode shape is similar for all six physical models (Figure 6). There is some digression from this shape in the Stage 2-int model (site 9,14,15). This is possibly due to inadequate fixation of the response accelerometer using the beeswax at these sites which led to lower than expected response magnitudes. Nevertheless the maximum and minimum of the fundamental mode shape are consistent for all six models. The similarity in the mode shape across models despite their different interface conditions supports the

suggestion that this may not be the optimum mode to investigate the detection of OI progression.

By contrast, the third frequency mode shape reveals differences between the models. In particular the Stage 2 and Stage 3 models have similar mode shapes along most of the model length while the Stage 1 and Stage 4 models exhibit differences in both shape and magnitude. The results suggest that monitoring the mode shapes could be a complementary or alternative method to monitoring the change in the resonant frequency values and could be used to identify changes at the femur-implant interface. Previously, mode shape analysis was not conducted by Xu and Shao (Shao, Xu et al. 2007; Xu, Shao et al. 2005) because their methodology did not allow for it. Therefore the modal analysis technique used in the current study has enabled further investigation of the feasibility of vibration analysis applied to the TFOI system.

The physical models with their different femur-implant interface conditions were developed to simulate key stages of OI progression based on histological observations of titanium implants in long bones and the mechanical properties of the bone-titanium interface. However the physical models cannot replicate the complex mechano-biological *in vivo* system in full. This is a limitation of using physical models to represent a biological system. Nevertheless the current study provides an iterative improvement in the physical modeling of the TFOI system. Furthermore using physical models enables all other variables of the system to be controlled. This allows the vibration methodology to be evaluated with respect to detecting femur-implant interfacial changes only. This is not possible using *in vitro* or *in vivo* models.

The cantilevered resin block was used as the femoral head boundary condition. The authors acknowledge that the *in vivo* femoral boundary condition is more complex than this. However the resin block was considered an appropriate first estimation of the boundary condition *in vivo*. Furthermore a similar boundary condition has been used successfully in the modal analysis of the tibia (Nikiforidis, Bezerianos et al. 1990).

5. CONCLUSION

A series of physical models were developed to simulate the mechanical characteristics of key stages in OI progression. Changes in the resonant frequencies and mode shapes as a result of physical property changes at the femur-implant interface were demonstrated, showing that the modal analysis technique is capable of detecting the incremental interfacial changes.

The findings indicate that higher resonances and their mode shapes may be more appropriate for the detection of OI progression than the fundamental resonance. The model boundary conditions may affect the success of the modal analysis technique and further investigation of boundary condition influence is required.

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