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Title: A clinically applicable non-invasive method to quantitatively assess the visco-hyperelastic properties of human heel pad with implications for assessing the risk of mechanical trauma.

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Abstract:

Introduction:

Pathological conditions such as diabetic foot and plantar heel pain are associated with changes in the mechanical properties of plantar soft tissue. However, the causes and implications of these changes are not yet fully understood. This is mainly because accurate assessment of the mechanical properties of plantar soft tissue in the clinic remains extremely challenging.

Purpose: To develop a clinically viable non-invasive method of assessing the mechanical properties of the heel pad. Furthermore the effect of non-linear mechanical behaviour of the heel pad on its ability to uniformly distribute foot-ground contact loads in light of the effect of overloading is also investigated.

Methods:

An automated custom device for ultrasound indentation was developed along with custom algorithms for the automated subject specific modeling of heel pad. Non-time-dependent and time-dependent material properties were inverse engineered from results from quasi-static indentation and stress relaxation test respectively. The validity of the calculated coefficients was assessed for five healthy participants. The implications of altered mechanical properties on the heel pad's ability to uniformly distribute plantar loading were also investigated in a parametric analysis.

Results:

The subject specific heel pad models with coefficients calculated based on quasi-static indentation and stress relaxation were able to accurately simulate dynamic indentation. Average error in the predicted forces for maximum deformation was only 6.6 ± 4.0%. When

the inverse engineered coefficients were used to simulate the first instance of heel strike the error in terms of peak plantar pressure was 27%. The parametric analysis indicated that the heel pad's ability to uniformly distribute plantar loads is influenced both by its overall deformability and by its stress/ strain behaviour. When overall deformability stays constant, changes in stress/strain behaviour leading to a more "linear" mechanical behaviour appear to improve the heel pad's ability to uniformly distribute plantar loading.

Conclusions:

The developed technique can accurately assess the visco-hyperelastic behaviour of heel pad. It was observed that specific change in stress-strain behaviour can enhance/weaken the heel pad's ability to uniformly distribute plantar loading that will increase/decrease the risk for overloading and trauma.

1. Introduction:

There is an increasing awareness that pathological conditions such as plantar heel pain or diabetic foot are associated with changes in the mechanical behaviour of plantar soft tissue (Hsu et al., 2007, 2000, 2002; Klaesner et al., 2002; Spears and Miller-Young, 2006; Zheng YP, Choi YK et al., 2000). Therefore, quantifying the mechanical properties of plantar soft tissue has been an exciting and evolving topic for the past few years. Several approaches involving *in vivo* and *in vitro* testing have been utilised to measure and quantify the subject specific mechanical behaviour of the plantar soft tissue (Chatzistergos et al., 2015; Erdemir et al., 2006; Miller-young et al., 2002; Naemi et al., 2016; Natali A.N., Fontanella C.G., 2012; Pai and Ledoux, 2010).

One of the most common techniques for the *in vivo* assessment of the mechanical behaviour of plantar soft tissue is ultrasound indentation. In this type of test, a load cell is used in series with an ultrasound probe enabling the simultaneous measurement of externally applied force and the internal deformations of plantar soft tissue (Chatzistergos et al., 2015; Erdemir et al., 2006).

Previous studies involving age-matched groups of people with diabetes (type-2) and people without diabetes have indicated that plantar soft tissue tends to be stiffer (Chao et al., 2011; Klaesner et al., 2002) and harder (Piaggesi et al., 1999) in diabetic patients. However, no real consensus has yet been reached in terms of the actual effect of diabetes on plantar soft tissue biomechanics as well as on the aetiology and possible implications of such changes.

Indeed existing results based on *in vivo* investigations vary significantly between studies even in the case of populations with no-known musculoskeletal pathology or diabetes (Spears and Miller-Young, 2006). This scattering of results can significantly compromise the value of ultrasound indentation data in comparative studies between different populations. A numerical study by Spears et al. highlighted that, tissue thickness, loading rate as well as the indenter geometry affect the force-deformation data of indentation tests and produce misleading results (Spears and Miller-Young, 2006).

Besides the inherent scattering of *in vivo* results between different studies, the comparison between these studies is also challenging because of differences in output measures. Although a number of studies have assessed tissue stiffness, differences in testing protocol (e.g. different loading rates, maximum applied loads etc.), and the differences in testing set-up (i.e. use of indenters with different footprint) make direct comparison between studies extremely challenging.

Quantifying the mechanical behaviour of the plantar soft tissue utilising the stress-strain behaviour rather than the force-deformation behaviour of the tissue can minimise the effects of tissue geometry, tissue thickness and the effect of indenter footprint area. Moreover, assessment of the time-dependent mechanical properties of plantar soft tissue (viscous properties) along with the non-time-dependent ones (hyperelastic properties) can eliminate the effect of loading rate on the results of the study.

Currently there is no direct assessment method for the non-invasive measurement of stress-strain behaviour of tissues (Atlas et al., 2009). At the same time the complex geometry and nonlinear mechanical properties of the plantar soft tissue does not allow any analytical solution without major simplification (Atlas et al., 2008).

In order to produce an objective, non-labour intensive and robust calculation of the mechanical behaviour of the plantar soft tissue, Erdemir et al. combined the in vivo indentation test with finite element (FE) modelling (Erdemir et al., 2006). For this purpose, an axisymmetric model of the indentation test was used to inverse engineer the hyperelastic material coefficients of bulk heel pad tissue. In order to increase the subject specificity of the inverse engineering process, Chatzistegos et al. utilised the ultrasound images that were collected during the indentation test to reconstruct the geometry of the heel pad and to design subject specific 2D models of the indentation tests (Chatzistergos et al., 2015).

The viscoelastic nature of heel pad was not included in neither of the aforementioned studies. However, considering the dynamic nature of loading of plantar soft tissue it becomes clear that assessing its time-dependent mechanical properties along with the non-time dependent ones is critical for the complete assessment and quantification of its mechanical behaviour (Fontanella et al., 2012; Natali, a. N; Fontanella, C. G; Carniel, E et al., 2012).

In this context, the aim of this study is to develop an accurate and clinically applicable method for the subject specific 3D modelling of the heel pad and the calculation of its visco-hyper-elastic material coefficients. The implications of altered tissue properties in the heel pad's ability to uniformly distribute external loads was also investigated in a parametric numerical analysis.

2. Method:

2.1 *In-vivo* testing

Five healthy participants with average age of 31±6y and average body mass of 71±15kg were recruited for this study after necessary ethical approval by the University's ethics committee (Table 1). All participants provided full informed consent prior to testing.

A custom-made automated ultrasound indentation device was utilised to perform quasi-static indentation, dynamic indentation and stress relaxation tests on the apex of the calcaneus (Figure 1a). The device consists of an ultrasound probe which is in series with a load cell (Zemic load cell, L6E, C3). The instrumented probe was attached on a rigid frame that is equipped with a linear actuator and a controller (TRM International Ltd., UK) to automate the movement of the probe. The device can generate linear movement at speeds between 0.005 mm/s and 52 mm/s to a predefined target-displacement or target-force. During testing, the imposed force and indenter displacement were recorded at 100 Hz while ultrasound images were recorded at 27 Hz. Synchronisation between the data from the indentation device and the ultrasound images was achieved with the help of a periodic pulse that is generated by the indentation device and imported into the ultrasound unit using an inbuilt input (ECG) port.

Testing was performed using a 10 MHz linear array ultrasound probe (LA523E, Esaote, Italy) with a rectangular footprint area of 60mm*10mm, which was adequately wide to image the entire width of the calcaneus. Before testing, the participants' left foot was fixed on the indentation device. The plantar surface of the participants' heel was covered with ultrasound gel and the probe was aligned in the anterior-posterior direction to image the apex of the calcaneus in sagittal plane (Figure 1b). Then the probe was rotated 90 degrees in the medio-lateral direction in order to image the tissue in frontal plane. The purpose of imaging the heel in the sagittal and frontal plane was to enable the reconstruction of its 3D shape. Finally, the ultrasound probe was pulled backwards to identify the point of first contact between the probe and the plantar surface of the heel and to measure initial heel pad thickness.

Dynamic and quasi-static tests were performed to the same subject specific target-load, based on preliminary pressure measurements, while stress relaxation to subject specific target-deformation. All tests were performed with the ultrasound probe imaging the medio-lateral (frontal) plane. Plantar pressure sensors (Matscan Walkway, Tekscan, Boston, MA, US) were used to measure net force during standing on an area equal to the footprint area of the ultrasound probe. Three independent measurements were recorded for 10 seconds at a sampling rate of 100 Hz and the average compressive load was calculated.

In dynamic testing, thirty load/ unload cycles were performed in total to the subject specific target-force with indenter displacement rate equal to 21 mm/s. The first 27 cycles were used for preconditioning and the last three to calculate an average force-deformation graph of the indentation test. The value of maximum indenter displacement for maximum force was also recorded for all thirty cycles to verify that the tissue was properly preconditioned.

Quasi-static tests were performed with loading rate equal to 0.052 mm/s to the same subject specific target-force as dynamic tests. Before testing, 27 preconditioning cycles were performed to minimise the effect of loading history to results with indenter displacement rate equal to 21 mm/s.

In the case of stress relaxation 27 load/ unload preconditioning cycles were followed by a sudden compression to 50% deformation with displacement rate equal to 41.6 mm/s. Deformation was then kept constant for one minute while force was recorded at 100 Hz. The aforementioned displacement rate is relevant to the mid-stance phase of gait and its value was decided based on the literature. More specifically, based on the in vivo heel pad deformations that were reported by Gefen and co-workers 2001 the deformation rate for mid-stance is calculated to be equal to 32±3 mm/s (Gefen et al., 2001).

In all cases, a safety deformation threshold of 80% compression was set to minimise the risk of injury. After the end of the test, and before releasing the participant's foot, heel width was also measured in the imaging plane using a digital calliper (Precision Gold). Tissue deformation (defined as the change of heel pad thickness) was measured after the completion of the test from ultrasound images using video analysis software (Kinovea, www.kinovea.org). After synchronisation, the recorded forces and calculated heel pad deformations were used to draw the force-deformation graph of the indentation test. In order to assess the repeatability of the process, testing was repeated for one of the participants and the results were compared.

The results from quasi-static indentation and stress relaxation were then used to inverse engineer the visco-hyper-elastic material coefficients of heel pad while the results from dynamic indentation were used for validation.

2.2 Inverse engineering:

Three dimensional (3D) subject specific FE models of the heel pad were designed using two ultrasound images which were recorded during the indentation test, namely: one frontal and one sagittal image of the heel at the area of the apex of the calcaneus (Figure 1b). A custom image processing algorithm (MATLAB and Statistics Toolbox Release 2014b, The

MathWorks, Inc., Natick, Massachusetts, US) was utilised to outline the boundaries of the calcaneus on both planes (Chatzistergos et al., 2015). The two, 2D curves representing the boundaries of the calcaneus on the frontal and sagittal plane were imported into Ansys (ANSYS® Academic Research, Release 16) and the 3D surface of the calcaneus was reconstructed by dragging the sagittal boundary along the frontal one. The initial thickness of heel pad model as well as its width were modified to correspond to the measured initial thickness and width of the heel respectively (Figure 2). The heel pad models were meshed using 13852 tetrahedral 4-noded elements and imported to FEBio software (Open source project, www.febio.org). Element size was decided through a sensitivity analysis to eliminate its effect on results.

In order to simulate the indentation test, the heel pad models were fixed at the calcaneus and the in vivo measured tissue deformation was imposed to the model of the ultrasound probe. The probe was simulated as a rigid body in contact with a bulk visco-hyperelastic soft tissue. The friction coefficient was set equal to zero to account for the effect of ultrasound gel (Erdemir et al., 2006).

The non-time-dependent mechanical behaviour of heel pad was simulated using the 1st order Ogden hyperelastic material model. The strain energy potential function for the Ogden material model (1st order) is defined as follows (Maas et al., 2015):

$$W(\lambda_1, \lambda_2, \lambda_3) = \frac{1}{2} c_p (J-1)^2 + \frac{C}{m^2} (\bar{\lambda}_1^m + \bar{\lambda}_2^m + \bar{\lambda}_3^m - 3 - mlnJ)$$

Where $\bar{\lambda}_i$ is the deviatoric principal stretch and c_p, C and m are material parameters.

Coefficients C and m are related to initial shear modulus and strain hardening respectively and c_p is directly related to the Poisson's ratio (υ) of the tissue. The heel pad was assumed to be nearly incompressible with Poisson's ratio close to 0.5 (υ=0.475).

The selection of this model was based on a preliminary analysis where 1st order Ogden was compared to two other commonly used models, namely Neo-Hookean and 1st order Mooney-Rivlin, and was proven to be able to simulate the mechanical behaviour of heel pad with satisfactory accuracy and the minimum number of coefficients (Appendix A).

Assuming the heel pad is nearly incompressible with Poisson's ratio equal to 0.475 leaves only two coefficients that need to be calculated, namely C and m. To calculate these coefficients, Levenberg-Marquardt optimisation algorithm (Maas et al., 2015) was employed to find the set of values that minimise the difference between the in-vivo measured quasi-static force-deformation graphs and the numerically calculated ones. The optimisation process was run three times with different random initial values in order to find the best sets of material coefficients for each participant. The objective function that was used in this minimisation problem was the sum of squared differences between predicted force values and experimental measured ones (Maas et al., 2015).

The viscoelastic material coefficients of the plantar soft tissue were estimated after the calculation of the hyperelastic mechanical coefficients based on the results of the stress-relaxation test. More specifically the time dependent mechanical behaviour of heel pad was simulated using the model shown below (Maas et al., 2015; Puso, M. A., and Weiss, 1998):

$$\boldsymbol{S}(t) = \int_{-\infty}^{t} G(t-s)\frac{d\boldsymbol{S}^{e}}{ds}ds,$$

Where *S(t)* is the Piola Kirchhoff stress tensor, $\boldsymbol{S}^e$ is the stress that relates to the elastic behaviour of the tissue, *G(t)* is the relaxation function of stress and *g*, *t* are the two material coefficients that need to be defined. Coefficient *g* is related to stain rate stiffening and *t* to relaxation time.

2.3 Validation:

The accuracy of this method was assessed by employing two different validation tests. For the first validation test, the subject specific models of the indentation tests with subject specific material coefficients were used to simulate dynamic indentation for all five participants. In these simulations, the models were loaded using the *in vivo* measured function of deformation over time. At the end, the numerically predicted reaction forces were compared against the experimentally measured values and their average percent error was calculated.

In the second validation test, numerical estimations of peak plantar pressure for the first instance of heel strike were compared against in vivo measurements. For this purpose, peak plantar pressure for the first instance of heel strike was measured for one participant and a detailed subject specific 3D model of their entire heel was designed based on MRI images (Figure 3). More specifically, the left foot of participant #4 (Table 1) was scanned using a 1.5 T MRI scanner and coronal T1 weighted 3D Fast Field Echo (FFE) images were recorded with in-plane and out of plane resolution of 0.23 mm and 1.00 mm respectively. The 3D geometry of the heel and of calcaneus was reconstructed using ScanIP (Simpleware, UK) and imported into Ansys (ANSYS® Academic Research, Release 16) for meshing. Using the previously inverse engineered subject specific material coefficients the final FE model was imported to FEBio to simulate heel strike. For this purpose, the model of the heel was fixed at an angle of 16 degrees (relative to a rigid surface simulating the ground) and loaded by applying a vertical compressive force to the ground (Figure 3). More specifically, 126N force was applied to the plate (i.e. the ground) in 0.01 second time span. Both the relative angle of heel and ground as well as the imposed loading were decided based on the in vivo measurements. Finally, the numerically predicted peak pressure for the first instance of heel strike was compared against the in vivo measured values and a percent error was calculated.

2.4 Parametric analysis

A parametric study was conducted to investigate the effect of hyperelastic and viscoelastic material coefficients on the ability of the tissue to evenly distribute loads. For this purpose, the aforementioned MRI-based model of heel strike was used to investigate the effect of altered tissue properties on peak pressure when the same external load was applied. Since there is no evidence on the extent of changes in mechanical properties of plantar soft tissue as a result of pathologies, a wide range of material has been explored. In the case of hyperelastic coefficients 238 scenarios were investigated in total, for C ranging between 25% and 500% of its reference values (C_{ref}) and m ranging between 25% and 300% of reference (m_{ref}) (Table 1). In the case of viscosity 100 scenarios were investigated in total for g and t values ranging between 25% and 200% of reference values (g_{ref}, t_{ref}) (Table 1). In all cases Poisson's ratio was kept equal to 0.475. In order to assess the effect of loading magnitude on the results of this analysis, selected scenarios were also investigated for externally applied force equal to 33%, 66% 133% of the references value.

3. Results:

3.1 In-vivo testing

Based on preliminary pressure measurements, the average subject specific target force for dynamic and quasi-static indentation was 46N ± 11N. The average heel pad thickness was 15mm ± 1mm which meant that the average subject specific target indenter displacement for during stress relaxation was equal to 7.5mm ± 0.5mm (Table 1).

Table 1: Key anthropometric characteristics of the participants (i.e. gender, age, body mass), parameters used to inform testing (i.e thickness of the heel pad, Subject specific target force) and key results from inverse engineering and validation. The material coefficients used as reference within the parametric analysis are presented underlined.

Participants	Gender	Age (y)	Body mass (kgr)	Heel pad thickness (mm)	Target force (N)	Deformation (mm) @ Target force in quasi-static test	C (kPa)	m	g	t	Validation error (%)
#1	M	23	74	14	35	4.9	53.97	18.48	0.75	2.2	11
#2	M	29	58	15	46	5.2	31.00	25.18	0.38	1.25	0.7
#3	M	34	90	16	45	6.2	22.25	29.61	0.63	5.5	9.0
#4	M	39	80	16	64	6.3	81.63	12.10	0.51	3.9	5.1
#5	F	29	55	15	41	6.0	16.55	33.00	0.72	5.25	7.0
Average		**31**	**71**	**15**	**46**	**5.7**	**41.48**	**23.68**	**0.65**	**3.62**	**6.6**
STDEV		**6**	**15**	**1**	**11**	**0.6**	**26.61**	**8.45**	**0.23**	**1.86**	**4.0**

Plotting maximum indenter displacement over the initial thickness of the heel pad for target force for thirty load/ unload cycles of dynamic indentation indicates that the mechanical behaviour of heel pad remains practically constant after the first 10 preconditioning cycles (Figure 4). This observation verifies that when measurements were taken (after 27 preconditioning cycles) the behaviour of the tissue remained constant.

The experimental results from *in vivo* tests highlighted the non-linear time dependent behaviour of the heel pad under compression (Figure 5). Test/ retest indicated that the testing process is repeatable. More specifically repeating the same test (i.e. dynamic indentation) for one participant in two different days gave differences of 6% and 2% in the maximum force and hysteresis respectively.

3.2 Inverse engineering:

Using 1st order Ogden hyperelastic material model to inverse engineer the hyperelastic coefficients for the remaining participants gave average values of C = 41.5 ± 26.6kPa and m = 23.7 ± 8.4 (Table 1). The respective average values for the viscoelastic coefficients were g= 0.645 ± 0.223 and t = 3.62 ± 1.86 (Table 1).

3.3 Validation:

The first validation test indicated that, using visco-hyperelastic material coefficients that were inverse engineered based on quasi-static indentation and stress relaxation enable an accurate simulation of dynamic indentation (Figure 5). The accuracy of dynamic loading simulation was assessed by comparing the numerically predicted maximum force that was calculated using the inverse engineered coefficients, against the in-vivo measured values for each participant (Table 1). The average absolute difference between the numerically calculated force for maximum deformation and the *in vivo* measured one was only 6.6 ±

4.0% (Table 1). For the second validation test, the difference between the numerically calculated peak plantar pressure and the in vivo measured one for the first instance of heel strike was 27% (i.e. 201 KPa and 158 KPa respectively).

3.4 Parametric analysis:

As shown in figure 6, the changes in the hyperelastic coefficients *C* and *m* can significantly affect peak plantar pressure for the same externally applied net force. Indeed, peak plantar pressure increases both with coefficient *C* and *m*. Increasing the value of *C* by 50% relatively to its reference value (Table 1), while keeping coefficient *m* constant, increases the peak plantar pressure by 6.7%, whereas decreasing *C* by 50%, decreases the peak plantar pressure by 7.1%.

Changing the strain stiffening coefficient (*m*) appears to affect peak plantar pressure more drastically. Indeed, increasing *m* by 50% increases the peak plantar pressure by 15.8% while decreasing *m* by 50% decreases peak plantar pressure by 7.2%. Increasing both the initial shear modulus and strain stiffening increase the peak plantar pressure severely.

Plotting peak plantar pressure over maximum deformation highlights the importance of the non-linear nature of the stress strain behaviour. As indicated in figure 7, the material properties of plantar soft tissue can change in ways that do not affect their overall deformation under the same load but still affect the capacity to uniformly distribute plantar loading. Focusing on the cases where overall deformation remains constant for the same external load indicates that more linear stress-strain behaviours can enhance the tissue's capability to uniformly distribute plantar loads. The compressive stress-strain graphs for three sets of coefficients that develop the same deformation (i.e. 6mm) under the same force are presented in Figure 7b. Peak plantar pressure underneath the heel with the more linear behaviour (C=250%C_{ref}, m= 99%m_{ref}) of the bulk soft tissue is 10% lower compared to the case with more nonlinear behaviour (C= C_{ref}, m=210% m_{ref}).

The effect of viscoelastic coefficients on the peak plantar pressure was also investigated. Increasing the viscoelastic stiffening coefficient value g to 125% of g_{ref}, decreases the peak plantar pressure by just 1 kPa. However, changing the relaxation time coefficient value (t), does not affect the peak plantar pressure in the first instance of heel strike, which was expected for the simulated loading scenario. Decreasing or increasing the magnitude of the externally applied force changed the values of maximum deformation and peak pressure, however the trend in terms of the relationship between material properties and pressure remained the same (Supplementary material Figure S1).

4. Discussion:

Pathological conditions such as diabetic foot and plantar heel pain are associated with changes in the mechanical properties of plantar soft tissue. However, the causes and implications of these changes are not yet fully understood. This is mainly because accurate assessment of the mechanical properties of plantar soft tissue in the clinic remains extremely challenging.

Diabetes is a growing health problem and diabetic foot is one of the main complications of diabetic patients (Boulton et al., 2005;). According to literature, 15% of people with diabetes world-wide will develop foot ulcer that can lead to amputation (Gordois et al., 2003). Although it is known that the risk of ulceration in certain areas of the foot is higher (i.e. heel, metatarsal head and hallux), the mechanism behind formation of ulcer is not yet fully understood ("Diabetes in the UK," 2011).

To address these challenges, the current study presents a methodology for the in vivo testing, the automated design of 3D subject specific models and the inverse engineering of the visco-hyper-elastic material coefficients of heel pad. For this purpose a series of mechanical tests was performed at the area of the apex of the calcaneus using an ultrasound indentation device. The inability to rigidly fix the calcaneus in vivo is considered as an inherent challenge of in vivo testing. This limitation can render unreliable calculations of deformation that are based on the displacement of the indenter (Chatzistergos et al., 2015; Erdemir et al., 2006; Naemi et al., 2016). The use of ultrasound imaging to directly measure heel pad deformations was deemed necessary and was implemented in this study.

Previous reports have established that human heel pad has visco-hyperelastic behaviour (Fontanella et al., 2012; Natali et al., 2010). Although different material models are available to describe the mechanical behaviour of the heel pad, the number of coefficients which can be inverse engineered from one test is limited. Therefore, the best material model is one that offers accuracy with the minimum number of coefficients. Initial investigation indicated that the 1st order Ogden hyperelastic material model can predict the non-time-dependent behaviour of heel pad with the least number of coefficients.

The importance of assessing the time-dependent aspects of the mechanical behaviour of heel pad has been highlighted in relevant literature (Fontanella et al., 2012; Kardeh et al., 2016). In order to quantitatively assess the mechanical behaviour of heel pad Fontanella et al. used a twelve coefficient visco-hyperelastic model. These coefficients were calculated using a combination of in-vivo data and data from literature (Fontanella et al., 2012). In another study by Kardeh et al. inverse engineered eight visco-hyperelastic coefficients form three in-vivo tests (Kardeh et al., 2016). In both cases the design of FE models was based on MRI scanning which was performed in addition to in vivo testing. In contrast to the aforementioned studies the method presented in the present paper is capable of producing accurate subject specific FE models and to inverse engineer the visco-hyperelastic material coefficients of plantar soft tissue only based on the data collected during two in vivo tests performed using a single device and with no need for time consuming and labour intensive geometry reconstruction.

More specifically, in the present study subject specific visco-hyperelastic mechanical properties of the heel pad of five healthy volunteers were calculated based on quasi-static indentation and stress relaxation tests. Erdemir et al. (Erdemir et al., 2006) reported the average initial shear modulus of the heel pad of 20 participants without diabetes as 16.54 kPa ±8.27 kPa. In the present study the average initial shear modulus for five participants can be calculated as 0.5*C= 20.74 kPa which is within the range reported by Erdemir and co-workers (Erdemir et al., 2006).

The accuracy of the subject specific FE models and of the inverse engineered coefficients was assessed for indentation scenarios different to those used for inverse engineering as well as for clinically relevant loading scenarios such as heel strike. The subject specific models of the in vivo tests were able to accurately simulate dynamic indentation and predict force for maximum deformation with an average error of 6.6 ± 4.0%. Moreover, using an MRI based model of the heel of one participant and subject specific coefficients to simulate heel strike enabled assessing peak plantar pressure with an error of 27%.

Investigating the effect of altered tissue properties on peak plantar pressure for the same value of plantar load indicated that increasing both initial shear modulus and strain stiffening can significantly affect the tissue's capacity to uniformly distribute plantar loading. However, the strain stiffening behaviour of heel pad appears to affect its ability to uniformly distribute plantar loads more severely compared to initial shear modulus.

Moreover, as demonstrated in figure 7, there are different sets of coefficients which can lead to the same deformation for the same compressive load; however, the stress-strain graph of each set is different. As a results of that, peak plantar pressure and the ability of tissue for distributing plantar load for each of the conditions is different. Even though overall deformability of the heel pad, as defined by maximum deformation for known force, clearly affects peak pressure, how ultimate deformation is reached plays also an important role. This finding quantifies for the first time the importance of the shape of the stress-strain graph of plantar soft tissue for its ability to fulfil its mechanical role namely, uniformly distribute plantar loads. Indeed, more linear stress-strain behaviour appears to promote uniform distribution of plantar loads more effectively compared to more non-linear one. This finding also indicates that assessing tissue stiffness based on the maximum deformation for known force to study plantar soft tissues' ability to uniformly distribute plantar loads could give misleading results. This needs to be considered in studies which quantify the stiffness of the soft tissue by measuring maximum force and maximum deformation without reflecting the tissue's nonlinear behaviour during loading (Chao et al., 2010; Yarnitzky et al., 2006).

Changes in the material properties that reduce the tissues' capacity to uniformly distribute plantar loads can make the tissue more vulnerable to overloading and mechanical trauma. Hence monitoring the possible changes in tissue biomechanics in time for people with feet at risk could possibly improve therapeutic outcome.

Even though some data on the differences between healthy and diabetic plantar soft tissue is available (Erdemir et al., 2006), the actual extent to which the properties of plantar soft tissue can change as a result of diabetes, internal trauma or other foot related pathologies is not yet explored. Because of this and in order to understand the relationship between heel pad mechanical properties and its ability to uniformly distribute plantar loads relatively wide range of coefficient's values was investigated. Further testing is required to identify the clinically relevant range of changes of plantar soft tissue.

For the purpose of this study, plantar soft tissue was assumed to be a uniform bulk material. This simplification of the tissue's internal structure means that the presented models are unlikely to be able to accurately predict internal stress and internal strain of the heel pad. For this reason the investigation presented here was limited to output measures related to the macroscopic behaviour of the heel pad such as overall deformation and peak plantar pressure. While the mechanical properties of the most heavily loaded part of the heel pad during the stance phase was investigated in this study, these properties may not represent the mechanical behaviour of the entire heel pad. Further testing at various sites is needed to assess the possible effect of regional differences in the mechanical properties of the heel pad.

In deed the modelling method presented in this study, was shown to offer satisfactory accuracy for the simulation and study of the macroscopic behaviour of the heel pad under quasi-static and dynamic loading. Additionally, the FE modelling procedure presented in this study, enables the calculation of the visco-hyperelastic mechanical properties of the heel pad. The clinical applicability of the presented technique is significantly enhanced compared to literature by the use of ultrasound imaging and by the fact the all post processing processes (e.g. geometry reconstruction, subject specific modelling etc.) can be completely automated (Behforootan, S; Chatzistergos, PE; Naemi, R; Chockalingam, n.d.). Although the method presented here was only employed on the heel, upon minor modifications it can be used for the other areas of the foot vulnerable to overloading and mechanical trauma such as metatarsal heads and hallux.

5. Conclusion:

The presented in vivo measurement-based technique can accurately assess the visco-hyperelastic material coefficients of heel pad. Moreover, it was observed that specific changes in the heel pad's stress-strain behaviour can enhance or weaken its ability to uniformly distribute plantar loading, thus increasing or decreasing, respectively, the risk for overloading and trauma.

Conflicts of interest: None

Appendix A:

The hyperelastic mechanical behaviour of heel pad was simulated using three different material models that are commonly used to model planar soft tissue and their ability of each material model to simulate the indentation test was assessed. The material models which were included in this comparison are: Neo-Hookean, 1st order Mooney-Rivlin and 1st order Ogden.

The strain energy potential function for Neo-Hookean material model is defined as (Maas et al., 2015):

$$W(\lambda_1, \lambda_2, \lambda_3) = \frac{\mu}{2}(I_1 - 1)^2 + \mu \ln J + \frac{\lambda}{2}(\ln J)^2$$

Where μ and λ are the lame parameters from linear elasticity. I is the invariant of the deviatoric part of the right Cauchy-Green deformation tensor and J relates to the incompressibility. This model reduces to the isotropic linear elastic model for small strains and rotations.

The strain energy function for Mooney-Rivlin material model is defined as:

$$W(\lambda_1, \lambda_2, \lambda_3) = C_1(\overline{I_1} - 3) + C_2(\overline{I_2} - 3) + \frac{1}{2}\mathrm{K}\,(\ln J)^2$$

Where the C_1 and C_2 are the Mooney-Rivlin material coefficients. The variables $\overline{I_1}$ and $\overline{I_2}$ are the first and second invariants of the deviatoric right Cauchy-Green deformation tensor. The coefficient K is a bulk modulus and J is the determinant of the deformation gradient tensor.

The strain energy potential function for Ogden material model is defined as (Maas et al., 2015):

$$W(\lambda_1, \lambda_2, \lambda_3) = \frac{1}{2} c_p (J-1)^2 + \frac{C}{m^2} (\bar{\lambda}_1^m + \bar{\lambda}_2^m + \bar{\lambda}_3^m - 3 - m_k lnJ)$$

Where $\overline{\lambda_i}$ is the deviatoric principal stretch and c_p, c_k and m_k are material parameters. Coefficients c_k and m_k are related to initial shear modulus and strain hardening respectively and c_p is indirectly related to the Poisson's ratio (υ) of the material. The heel was assumed nearly incompressible material with Poisson's ratio close to 0.5 (υ=0.475) thus the model was left with two unknown coefficients.

Considering limitation in term of the number of coefficients that can be inverse engineered from one test, the investigation was limited to first order models. Assuming the heel pad is nearly incompressible with Poisson's ratio equal to 0.475 leaves two coefficients that need to be calculated in the cases of 1st order Mooney-Rivlin and 1st order Ogden and one coefficient for the Neo-Hookean model. To calculate these coefficients, Levenberg-Marquardt optimisation algorithm (Maas et al., 2015) was employed to find the set of coefficients that minimise the difference between the in-vivo measured quasi-static force-deformation graphs and the numerically calculated ones. The optimisation program was run 10 times with random initial values for the coefficients in order to find the best sets of coefficients. The objective function that was used in this minimisation problem was the sum of squared differences between predicted force values and experimental measured ones. The minimum values for the objective function achieved using 1st order Ogden, 1st order Mooney-Rivlin and Neo-Hookean were 12, 644 and 840 respectively. Based on this, it was concluded that 1st order Ogden hyperelastic is the material model that can generate the best fit between numerical and in vivo results with the minimum number of coefficients.

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Figures:

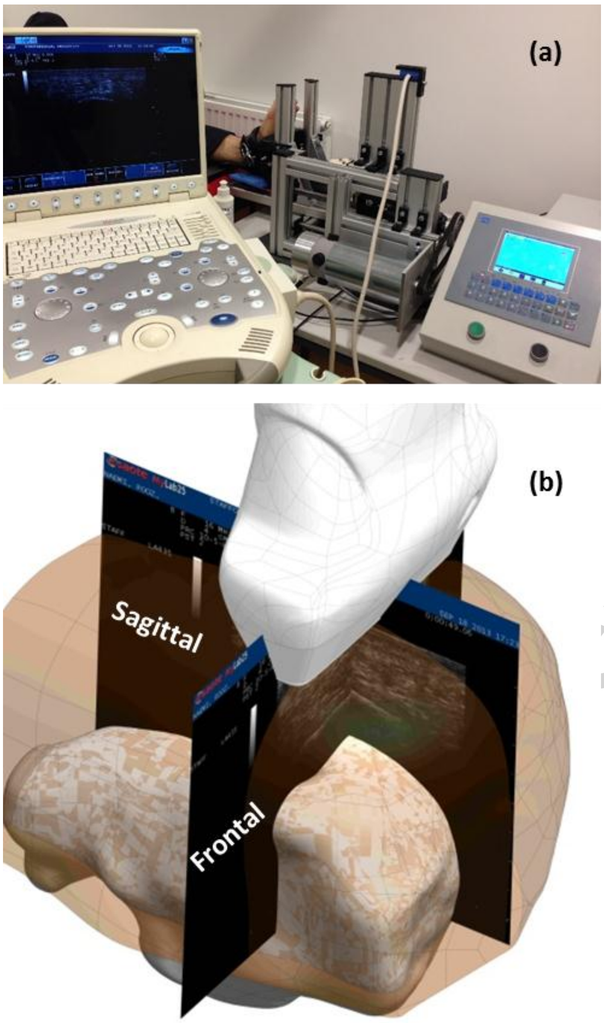

Figure 1. The custom automated ultrasound indentation device (a) and the imaging planes that were used to reconstruct the geometry of heel pad (b).

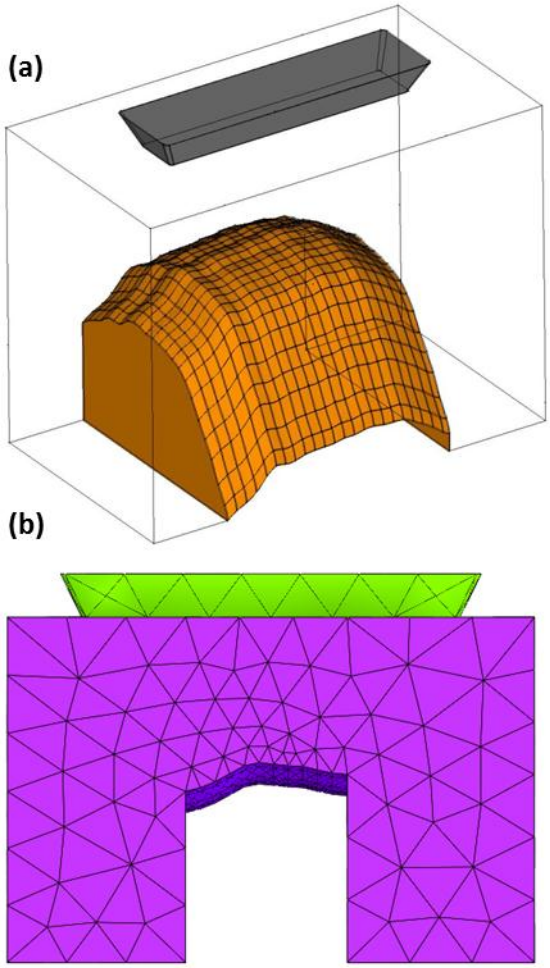

Figure 2. Example of the reconstructed surface of the calcaneus based on the ultrasound images recorded during the indentation test (a) and the final produced subject specific FE model (b).

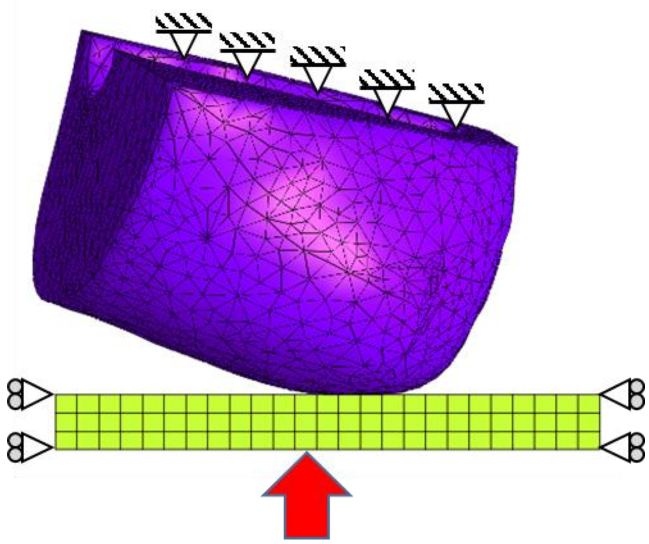

Figure 3. The MRI based 3D model of the heel.

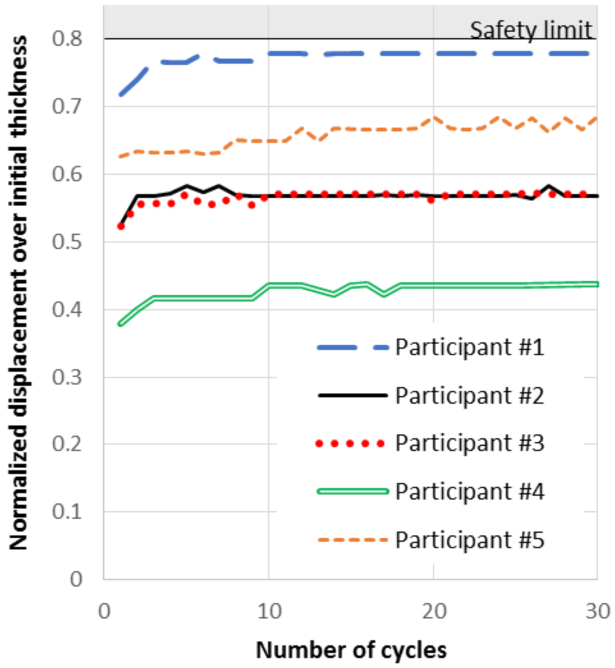

Figure 4. The effect of preconditioning for all participants as quantified by changes in maximum indenter displacement for force equal to the subject specific target force over 30 load/ unload cycles. The displacement of the indenter is presented normalised over the initial thickness of the heel pad at the area of the apex of calcaneus. The safety limit of 80% normalised indenter displacement that was used during testing is also shown.

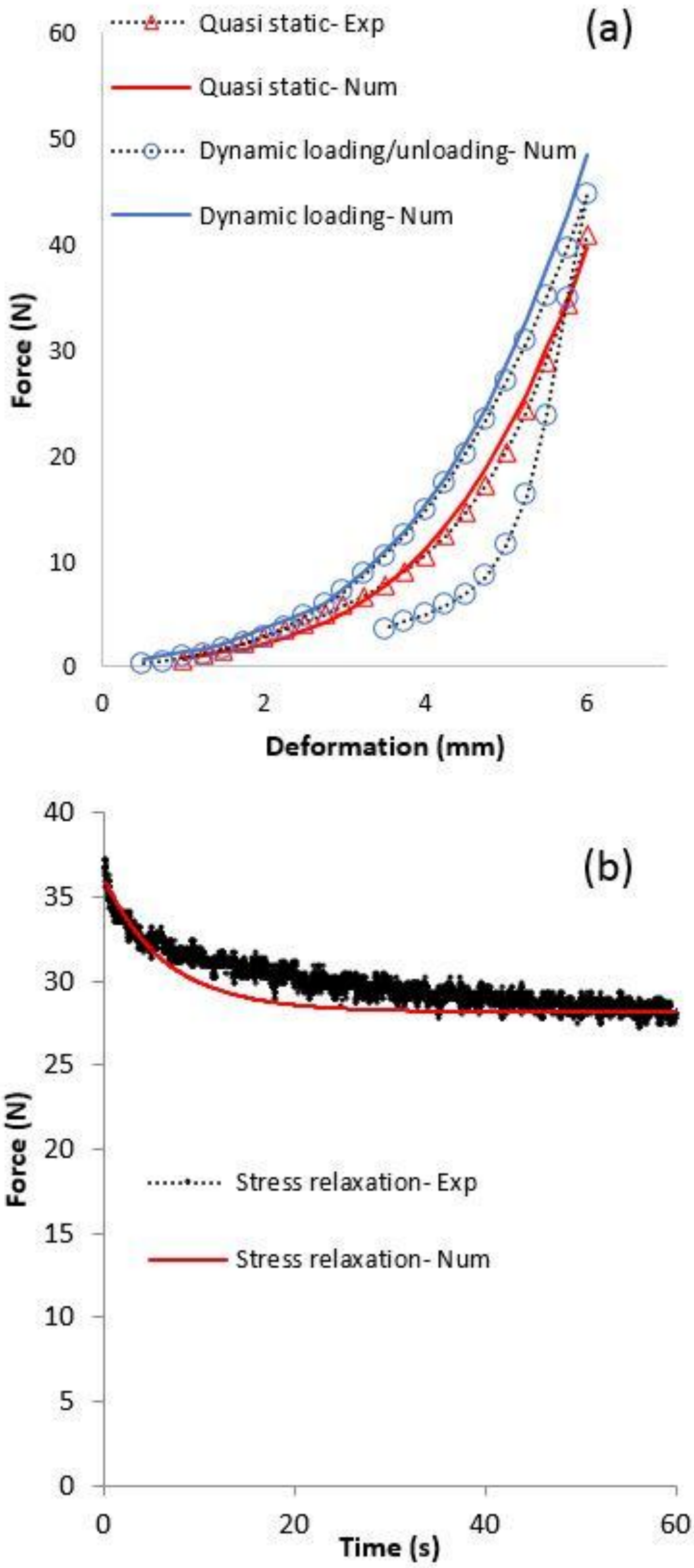

Figure 5. The typical experimental results from in-vivo tests and numerical results from finite element analysis: (a) quasi-static and dynamic indentation and (b) stress-relaxation.

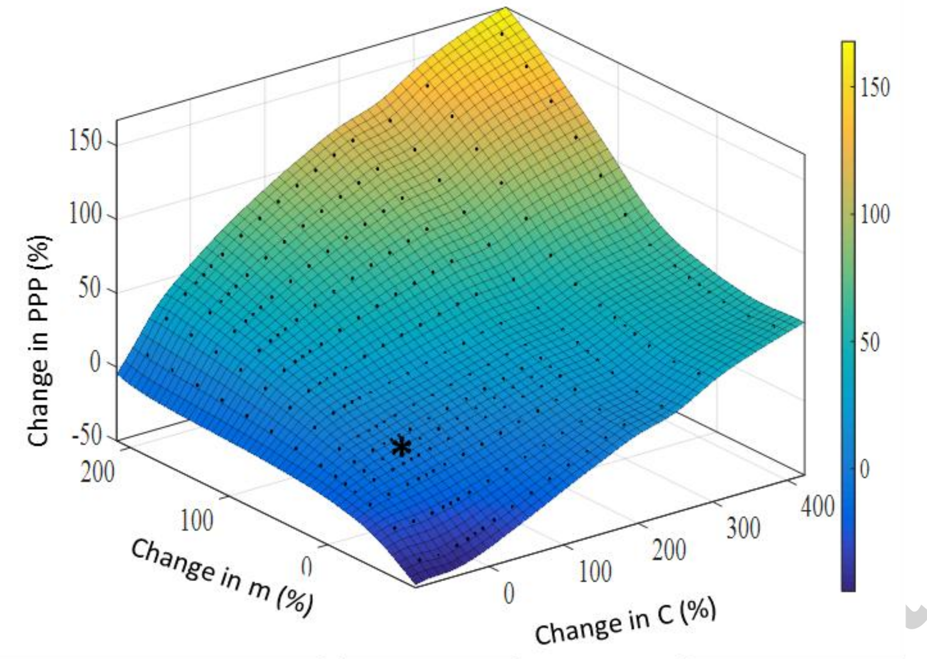

Figure 6. Effect of changes in the hyperelastic (Ogden 1st order) mechanical coefficients m and C on peak plantar pressure (PPP). The results for the reference values of hyperelastic confidents are marked with (*).

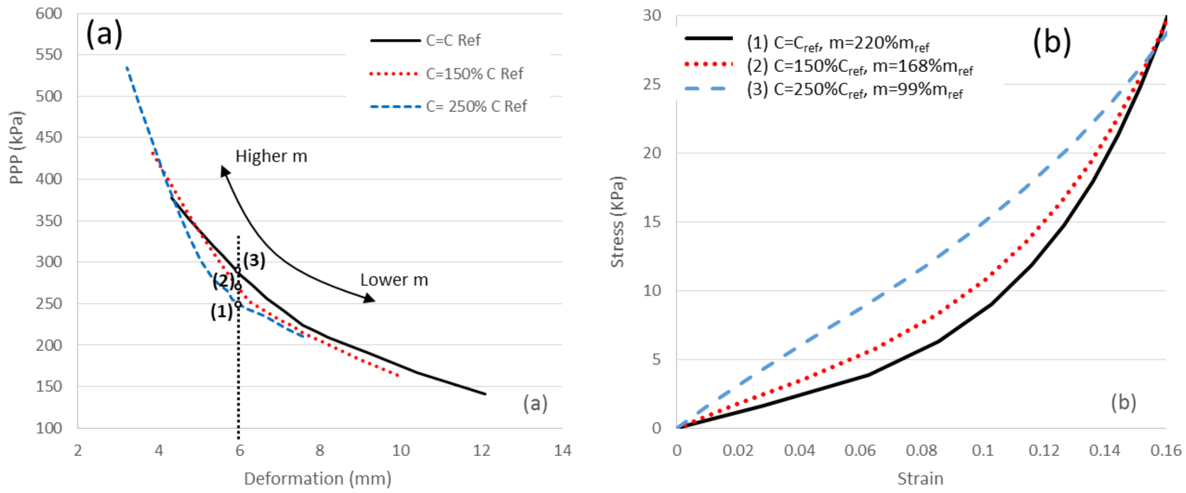

Figure 7. The effect of stress-strain behaviour of heel pad on its ability to uniformly distribute loading. (a) The relationship between peak plantar pressure (PPP) and deformation for different values of initial shear modulus (C) and strain stiffening (m). Three different pairs of values (i.e. (1)(2)(3)) are identified where the same amount of externally applied force leads to the same maximum deformation (i.e. 6 mm) but different peak pressure. (b) The numerically calculated compressive stress-strain graphs for materials with coefficients equal to the ones identified in graph 7a. In this case stress and strain correspond to the real stress and strain of ideal cubic samples of materials that are subjected to standardised compression. The values of the coefficients are also presented relative to the reference ones (C_{ref}, m_{ref}).

Highlights

- An automated subject specific modeling technique was developed
- Heel pad’s mechanical properties were inverse engineered for healthy participants
- The effect of visco-hyperelastic mechanical coefficients was investigated
- Deformability is not enough to assess tissue mechanical viability
- The effect of the shape of stress-strain graph was assessed for the first time