3D FOOT SHAPE RECONSTRUCTION FROM PLANTAR PRESSURE CARTOGRAPHIES

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ABSTRACT
Foot problems due to diabetics, age or joint disease are treated by application of personalized foot insoles. Designing an insole requires the knowledge of the plantar foot shape. Several techniques are used to get or estimate the 3D foot shape.
Among them, plantar pressure cartographies are used by podiatrists to estimate the design of the foot insoles. Unfortunately, due to the lack of conversion tool between pressure maps and foot shape, several iterations of design and measurements are needed.
In the present study, a new method aiming to extract the 3D foot shape under pressure is presented. The main elements of the foot that induce the plantar pressure distribution are presented in a specific biomechanical model. We present and discuss extraction of the 3D pressured foot shape from the barefoot plantar pressure measurements.

KEY WORDS
Foot shape reconstruction, biomechanical foot modeling, plantar pressure cartographies.

1. Introduction

The personalized foot insole is often used in decreasing foot pain due to diabetics, age or joint disease. The shape of an insole depends on the underfoot shape while the foot is under pressure.
By now, the design of an insole is estimated through the measurement of the foot shape using several techniques. The most popular one uses the plaster. The foot insole is modelled with respect to the shapes leaved in the plaster. This method is approximate because the foot shape is measured under an unknown applied force. 3D scanners [1] [2] provide also the foot shape but not under pressures. Others use a video reconstruction from multi camera [3-8]. These techniques are limited by the lack of knowledge on the plantar anatomical characteristics. Other techniques based on the plantar pressure measurements are used in designing compensating insoles. This design is rather empirical and requires several measurements with and without the insole to include its effect on the plantar compression.

In this work, we present a new method aiming to link the 3D foot shape to the plantar pressure cartography. It is to be used in the insole manufacturing and the simulation of the pressure repartition under feet. The extraction of the foot shape from the plantar pressure cartographies is detailed in the case of the standing position.

2. Background, foot model

The reconstruction of the foot shape requires the knowledge of the main foot characteristics that induce the pressure repartition.
Many researchers focus on the biomechanical foot model. One can mention 3D finite element models of the foot and ankle which have been developed to study ulcer formation [9] [10], or to predict the forces between the foot and different types of support, such as moulded insoles, varying insole stiffness [11-14]. Models of coupled foot-boot [15] have also been proposed.
This modelling technique requires prior knowledge of the foot shape in 3D, provided by MRI measurements, CT scanners or video reconstruction. Unfortunately, most podiatrists do not have these expensive and sophisticated devices at their disposal.
With the aims to offer the podiatrist a practical tool to extract the 3D foot shape and the ability to simulate plantar pressure distribution, we have derived a new model of the foot, taking into account the foot shape. The model must also take into account the properties of the shoe insole. Of course this model will have to be efficient with the only data that podiatrists can measure on their own.
The proposed original foot model describes the relationship between the force applied to the foot and the plantar pressure distribution via the 3D foot shape and the elastic material properties. The plantar pressure repartition depends on:

- The foot seat (FS) describing the foot position in the space.
- The internal foot shape (IS) defining the non compressive material of foot.
- The characteristics of the elastic medium (EM) describing the soft tissues compressive material.

The mathematical model parameters of each element are foot dependent.

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2.1 Foot seat

The origin of the internal coordinates of the foot is the ankle, x being the lateral coordinate axis, y the longitudinal one (heel to first metatarsus) and z the vertical one.

The location of the foot in space is given by the vertical position of the ankle and the rotations around the ankle, as shown in Figure 1. The equation of the "foot seat" FS plan is:

$$z_{FS}(x,y) = \tan(\alpha)\cdot x + \tan(\beta)\cdot y + c = a\cdot x + b\cdot y + c$$

The terms $a\cdot x$ and $b\cdot y$ respectively describe the lateral and longitudinal rotations around the ankle, whereas c is the vertical displacement of the ankle.

The considered rotation angles are small. In the approximation of small angles $a = \tan(\alpha) = \alpha$ and $b = \tan(\beta) = \beta$.

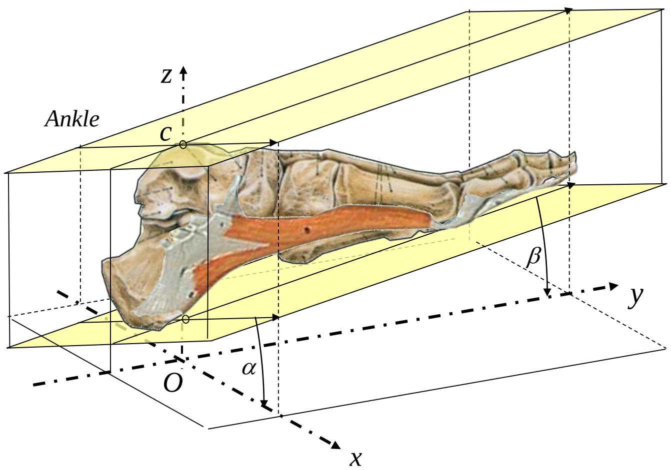

Figure 1: Illustration of the foot seat.

2.2 Internal foot shape

In order to distinguish the non compressive material (skeleton, ligament) from the compressive one (soft tissues), we define the Internal foot Shape IS, the surface profile describing the non-compressive material due to the skeleton covered by the other rigid media, such as the ligaments, as shown in Figure 2 A and 2 B.

The internal foot shape is described by the distance of each foot point to the foot seat plan. Since the rotation angles are small, the distance has its main component, $z_{IS}(x,y)$ along the z axis. The components along the x and y axes due to the rotations are neglected.

2.3 Elastic medium

The internal foot shape is covered by soft tissues as shown in Figure 3 C. The thickness and the elasticity of the soft tissues depends on the underfoot location. To simplify the modelling, it is assumed that the soft tissues can be described realistically by an equivalent uniform elastic medium EM. The elastic medium is uniform in terms of thickness and elastic properties, Figure 3 D.

The relation between pressure and compression of the EM should of course verify that there is no force when the foot is not in contact with the floor and that the EM cannot be compressed beyond its thickness. The vertical compression $z_C(x,y)$ of the EM is then directly related to the FS and IS functions. With contact, the compression becomes $z_C(x,y) = z_{FS}(x,y) + z_{IS}(x,y)$.

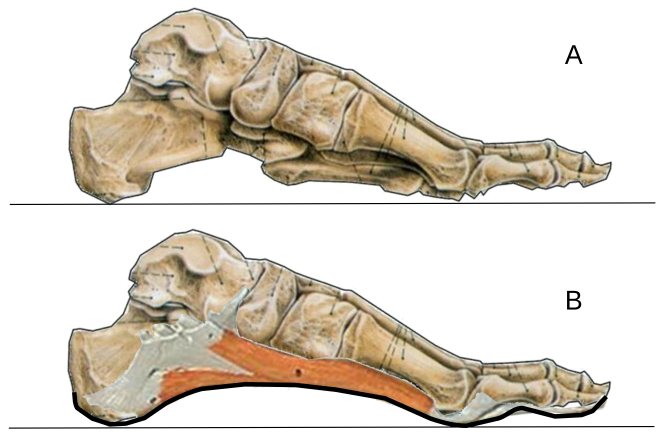

Figure 2: Foot model. A foot skeleton [16], B internal foot shape.

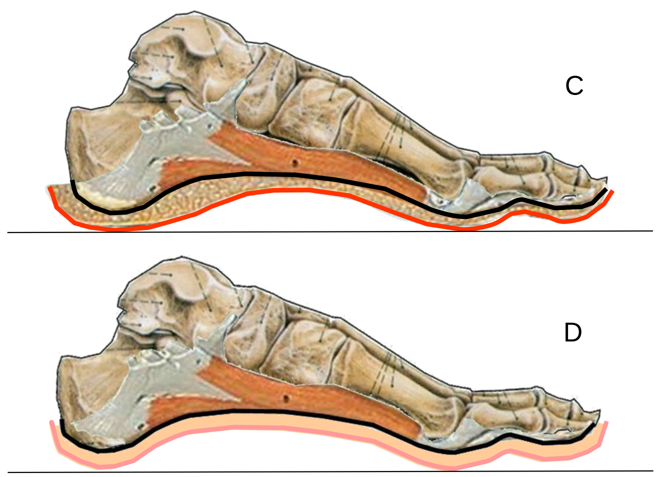

Figure 3: Elastic medium: C in red, soft tissues, D in pink, modelled elastic medium.

We experimentally verified that the function which links an elastic medium compression to the applied force can be described by:

$z_C = thick\cdot\left(1-\exp\left(-\dfrac{F}{stiff}\right)\right)$ for $F \geq 0$ and its inverse function $F = -stiff\cdot\ln\left(1-\dfrac{z_C}{thick}\right)$ for $0 \leq z_C < thick$.

The compression z_C (mm) depends on $thick$ (mm), the thickness, and $stiff$ (N / cm²), the stiffness. It varies with F (N / cm²) the pressure.

Having presented the main elements of the model, we now detail an application of this model, aiming to build the 3D foot shape from plantar pressure cartographies. Measurement of pressures will be made by using instrumented insoles.

3. Material

The plantar pressure cartographies under each foot are recorded by using the popular F-Scan Mobile® system (from Tekscan®) [17]. The posture has to vary during recording to be used as database to the numerical methods. This variation is natural during walking, but in standing position the subject is asked to exaggerate its posture in mediolateral and anteroposterior directions in order to get enough displacement amplitude to cover the entire surface of the foot. The problem is then time and spatially discretised. The size of a pixel corresponds to the F-Scan one, $S_p = 0.508 \times 0.508\, cm^2$.

4. Method

The pressure measurements are made with F-SCAN, at a frequency of 50 scans per second. From measurements one can obtain a set of pressure maps for different foot seats.

At each time step $i = 1 \ldots M$, for each pixel $k = 1 \ldots N$ of the foot, the force F_i^k is measured and stored. Note that each time step corresponds naturally to a different foot seat. The compression $z_{Ci}^{\ k}$ depends on two parameters: the seat plan $z_{FSi}^{\ k} = a_i \cdot x_k + b_i \cdot y_k + c_i$ and the internal foot shape z_{ISk}.

$$a_i \cdot x_k + b_i \cdot y_k + c_i + z_{IS}^{\ k} = thick \cdot \left(1 - \exp\left(-\frac{F_i^k}{stiff}\right)\right)$$

The unknown spatial parameters are $\left(a_i\,; b_i\,; c_i\,; z_{IS}^{\ k}\right)$. The relation shows that the thickness $thick$ appears as a proportional factor for those parameters. In the present case there is no possibility to determine the thickness from the relation. So it is fixed to an arbitrary realistic value acting as a proportional factor on the foot seat and the internal foot shape. The unknowns are reduced to the stiffness, $stiff$, the 3 foot seat parameters $\left(a_i\,; b_i\,; c_i\right)$ for each step i and the internal foot shape $z_{IS}^{\ k}$ for each pixel.

The total number of data $N \times M$, is larger than the total unknown values $N + 3 \times M$. The number N of pixels is about 500, and the number M of postures (pressure maps) is in the order of 1000 or more, depending on the duration of the measurement.

Let be E the summation of the square of the errors made on the compression, which can be written:

$$E = \sum_{i=1}^{M} \sum_{k=1}^{N} \left(thick \cdot \left(1 - \exp\left(-\frac{F_i^k}{stiff}\right)\right) - \left(a_i \cdot x_k + b_i \cdot y_k + c_i + z_{Pk}\right) \right)^2$$

Where the F_i^k are the experimental data.

Since thickness is fixed, at the minimum value of E, the values of the unknowns can be classically obtained by expressing that the partial derivatives of E are null with respect to each unknown:

$\forall i$: $\frac{\partial E}{\partial a_i} = 0$; $\frac{\partial E}{\partial b_i} = 0$; $\frac{\partial E}{\partial c_i} = 0$ and $\forall k$: $\frac{\partial E}{\partial z_{Pk}} = 0$ to which $\frac{\partial E}{\partial stiff} = 0$ should be added.

However, $\frac{\partial E}{\partial stiff} = 0$ do not lead to a linear form. So, different values of $stiff$ are tried within a dichotomy type method to determine which one minimises E. The other partial derivatives, $\left(\frac{\partial E}{\partial a_i} = \frac{\partial E}{\partial b_i} = \frac{\partial E}{\partial c_i} = \frac{\partial E}{\partial z_{Pk}} = 0\right)$, lead to a classical linear equation system which is solved for each of the stiffness fixed values.

The unknown parameters are obtained by minimizing the error E. The numerical method has been developed with MATLAB software on an ordinary desktop personal computer. The computing time is about 10 seconds.

5. Results

In the following, we present a measurement of the plantar pressure distribution, on a 70 kg male subject with a foot length of 26 cm, bare-footed, in a standing position. The F-SCAN pressure maps are recorded at a frequency of 50 frames/ second during 1 minute (3000 scans).

The calculated internal foot shape is shown in Figure 4. As exposed before, the height scale depends on the thickness of the elastic medium. For this measurement, it has been estimated to 5 mm. The foot shape height varies then between 1 and 4.5 mm. A variation of 2 mm can be observed between the mid-foot and the forefoot or the heel. Figure 5 presents the calculated variance. The variance includes implicit errors due to the simplified model hypothesis and measurement uncertainty. The root mean square error is less than 0.15 mm for the main support zones, which corresponds to 3 % of the maximum height.

Other high variance points may appear in the zones where only a few pressure data are available. In any case the precision is less than a few percent.

6. Discussion

The internal foot shape is extracted from the plantar pressure cartographies while the subject is standing. In this case, both feet support the real total weight of the subject, in contrast to results obtained when the podiatrist applies pressure only on one foot after another.

Under pressure, the compression of the elastic medium is taken into account in the mathematical calculation. Unlike the measurement made by using the plaster and the scanner where the foot is not under the pressure of the patient, the internal foot shape issued from plantar pressure cartographies depends on the patient weight and posture. The extraction of the foot shape from the plantar pressure maps, exhibits a variance about a few percents under the heel and the mid foot. A more important variance can be seen under the toes of 12 % of the height measured under the toes. This variance is due to the relatively low pressure measured under the toes during calibration. A possible solution is to apply a specific pressure under the toes when making measurements.

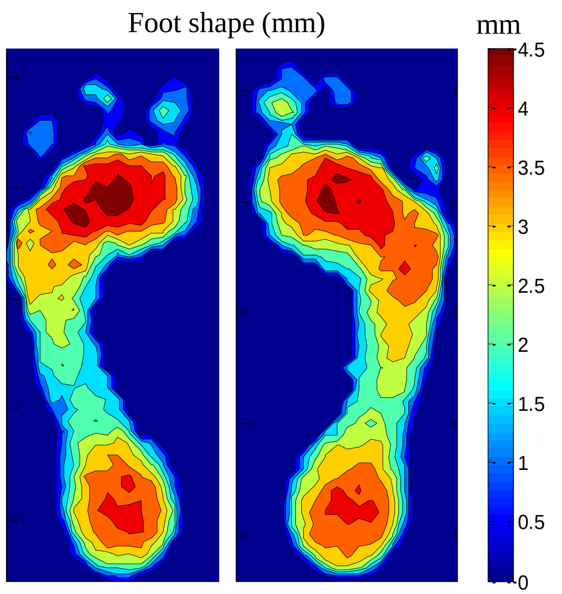

Figure 4: Internal foot shape altitude in mm.

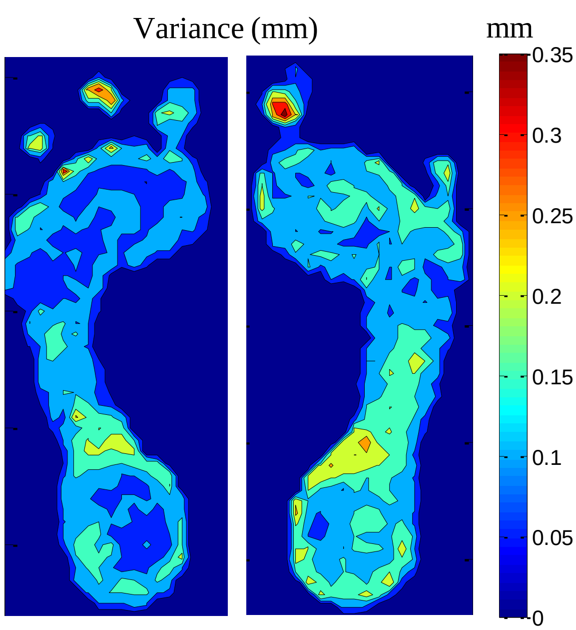

Figure 5: Internal foot shape variance in mm.

Starting from these measurements, the height of the extracted foot shape is extracted for a selected thickness of 5 mm. This thickness is comparable to the thickness of the plantar soft tissue. Otherwise, when the patient is standing on an elastomeric insole during measurements, we can adjust the total thickness value, depending on the stiffness of each medium and mechanical combination law.

The foot model is proposed here with hypothesis on the elastic medium characteristics. The low variance on the foot shape extraction confirms our choice on the parameters assumptions of the foot model by linking the force applied on the foot to the pressure distribution via a uniform elastic medium. This approximation of a uniform elastic medium in place of heterogeneous soft tissues minimizes the complexity of calculation.

We also verified the validity of the method during walking: the plantar pressures have been recorded and the internal foot shape extracted for the same subject. Results show a high similarity in the IS extracted in standing position and during walking. Therefore, using the plantar pressure to extract the foot shape during walking permits to refine the design of insoles depending on the gait.

As exposed before, it should be noted that using a compensated insole during the measure of the plantar pressure changes the IS. In this case, the calculated IS include the form of the compensated insoles and the IS calculated when bare foot.

7. Conclusion

We have proposed a new method to extract the 3D internal foot shape by using a biomechanical foot model based on simplified anatomical modeling of the human foot. The 3D internal foot shape under pressure is calculated from plantar pressure cartographies, issued from usual matrix systems, such as F-Scan, avoiding the iterative techniques.

This new method provides a practical tool to design personalized foot insoles for patients. In the future, this model could be further developed to simulate the action of added insoles on the underfoot pressure when assigning different thickness and stiffness to each area of the elastic medium.

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