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Original Article

Influence of insole slope on bone joint stress, foot bone stress, and foot pressure distribution

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Abstract

This study aimed to investigate the effect of insole slope on bone joint stress, foot bone stress, and foot pressure distribution by means of finite element (FE) methods. A three-dimensional (3D) FE model was reconstructed from computed tomography data using image processing and computer aided design software. A model of an insole placed under the foot skin in 3D FE model geometry was based on a Smile Feet insole. Four conditions for the analysis included one barefoot biomechanical analysis and three biomechanical analyses with 0-, 5-, and 10-degree insole slopes. In the barefoot analysis, high foot pressure was concentrated on the forefoot and heel regions. The insole redistributed the foot pressure to the mid-foot region and reduced stress at the intervertebral discs of L4-L5 and L5-S1, and at the hip joint. The slope of the in-sole resulted in changes in the stress exhibited in various joints and bones. Higher insole slopes tended to increase the stress, especially at the ankle joint and foot bones.

Keywords: insole slope, bone stress, joint stress, foot pressure

1. Introduction

Insole is an orthotic device placed under a foot and one of its major functions is to redistribute foot pressure to prevent ulceration. It is used especially in diabetic patients with peripheral neuropathy (Lipsitz *et al*., 2015), and vascular disease (Luo, Houston, Garbarini, Beattie, & Thongpop, 2011), as well as in patients with malalignment of the limb axis (Liu & Zhang, 2013). Various insole designs have been invented which aim to be used in a variety of functions. The first example is the custom-made insole. Its shape is fabricated according to the individual's foot geometry to provide accommodating support and reduce high pressure under the heel and metatarsal regions (Chen, Ju, & Tang, 2003). The second example is the heel-evaluated insole. It is used to restore the patient's gait to a symmetrical pattern (Tang *et al*., 2003). The third example is the wedge shape insole which provides a slope in the lateral-medial direction and is used for limb alignment correction (Liu & Zhang, 2013).

Current investigations for the development of insole designs focus on the material and shape. The insole materials have tended to move to elastic materials such as porous foam rubber containing ethylene vinyl acetate (EVA) (Tatàno *et al*., 2012), and layer-composites (Ghassemi, Mossayebi, Jam shidi, Naemi, & Karimi, 2015). Apart from the materials, the insole shape aims to reduce foot pressure. Many publications have reported their investigations of foot pressure related to

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insole design (Hellstrand Tang *et al*., 2014; Ibrahim, El-Hilaly, Taher, & Morsy, 2013; Liu & Zhang, 2013). In custom-made insole production, the shape of the insole can be designed according to the foot geometry to ensure uniform contact of the foot to the insole. Foot pressure is significantly reduced and redistributed uniformly throughout the entire foot (Chen *et al*., 2003). Although a custom-made insole is the best fit in terms of the clinical aspect, fabricated insoles can be done only one at a time. Some manufacturers prefer to fabricate insoles via mass production. In mass production, insoles are made in the same shape with a molding technique. Therefore, the shape of a mass produced insole cannot ensure total contact with the surface of the foot. As a result, the pattern of foot pressure distribution may concentrate in a high magnitude in some regions.

One interesting insole parameter which relates to the shape design is insole slope. Insole slope is associated with other parameters such as heel height which has drawn the attention of many authors (Ahmady, Soodmand, Soodmand & Milani, 2014; Chen, Tang, Hong, & Tang, 2015). However, these studies on insole have generally reported the performance of the insole by focusing only on the foot region. In addition, most studies related to insole are usually preformed experimentally using a pressure transducer to explore the foot pressure (Bus, Ulbrecht, & Cavanagh, 2004; Kato, Takada, Kawamura, Hotta, & Torii, 1996; Tang *et al*., 2014). Although foot pressure plays a critical role for insole design, understanding the effect of insole on the bones and joints may be useful for the design of better insoles to prevent progression of other diseases rather than only foot ulceration.

Among the mechanical parameters, stress can effectively describe the health conditions of bones and joints, for example higher stress on one knee compartment compared to the contralateral compartments may lead to osteoarthritis (Chantarapanich, Nanakorn, Chernchujit, & Sitthiseripratip, 2009). Although an *in vivo* investigation reflects the most-reliable data, implanting sensors to measure the stress at a bone joint is not possible and prohibited. In addition, *in vitro* tests, which are considered to be another effective testing approach, may be difficult to set up a realistic posture of bones and apply the correct boundary conditions.

As an alternative to *in vivo* and *in vitro* tests, the finite element (FE) method can be performed to obtain a numerical solution. The FE method allows a mechanical analysis to be performed on a computer. It provides an alternative to an analysis which has testing conditions that have difficulties in setting up and performing the experiment. The FE method has advantages, especially in the field of biomechanics. Most biomechanical experiments are complex and involve various organs and material properties. Most of the complex biomechanical experiments utilize FE for analysis. For example, Mahaisavariya, Chantarapanich, Rian suwan, and Sitthiseripratip (2014) performed a FE analysis to find the possibility of improving the surgical technique in trochanteric fracture with dynamic hip screws combined with an anterior-posterior buttress screw. Based on the FE method, all testing conditions and biomechanical analyses were performed on a computer. This study aims to utilize the FE method to investigate the effects of insole slope on joint stress, stress on foot bone structure, and foot pressure distribution.

2. Materials and Methods

All FE models were constructed using reverse engineering and medical imaging from computed tomography (CT) data. All analyses were performed using a FE software package (Marc Mentat, MSC Software Inc., USA) with multiple solver processes. The computer used in this study had an Intel Core i5-6200U processor running at 2.30 GHz with 4 GB of RAM memory. The average computational time was approximately 90 minutes per case.

Four FE analyses are employed that included one barefoot biomechanical analysis (without insole) and three biomechanical analyses with 0-, 5-, and 10-degree insole slopes. All FE models were created based on their analytical conditions and the following procedures.

2.1 Data acquisition and 3D model reconstructions

The anatomical data used in this study was digitized from the body of the first author (NC) using a 64-slice CT scanner. The region of interest for the digitization process ranged from the lumbar region toward the foot. The anatomical data was formatted into a Digital Imaging and Communications in Medicine (DICOM) file which could be imported into medical imaging software (Mimics, Materialise N.V., Belgium) for 3D Computer Aided Design (CAD) model construction. CT data makes it convenient to extract bone and peripheral skin geometries. The proper Hounsfield units were applied to extract bone structure and skin in the foot region. The 3D models obtained from the construction process using the CT data included 5 vertebrae of the lumbar spine (L1-L5), pelvis, femur, tibia, talus, foot bone structure, and soft tissue skin at the region of the feet. The lumbar facets and cartilages, which are invisible in CT images, were created using CAD software (VISI, Vero Software, UK) to complete the necessary geometries for FE analysis.

The 3D geometric model of the insole used in this study was a Smile Feet model received from Health Innovation & Design Limited Partnership. The model was adjusted for slopes of 0-, 5-, and 10- degrees.

Bone and skin models were aligned to the insole model. All final 3D geometric models were converted to triangular mesh models which were stored in stereolithography (.stl) file format.

2.2 FE model generations

All 3D triangular mesh models were converted into FE models by building up a four-node tetrahedron element based on their 3D topologies. The number of elements employed in this study ranged from 292,827 to 305,634 which corresponded to 79,395-82,616 nodes.

2.3 Material properties

All material properties assigned to FE models were assumed isotropic, homogenous, and linearly elastic. In the lumbar spine region, the lumbar bones were assigned an elastic modulus of 12,000 MPa and Poisson’s ratio of 0.3 (Dreischarf *et al*., 2014). The posterior bony elements were assigned an elastic modulus of 3,500 MPa and a Poisson’s ratio of 0.25 (Dreischarf *et al*., 2014). The pelvis bone was assigned the mechanical properties of 17,000 MPa for the elastic modulus and 0.28 for Poisson's ratio (Wei, Sun, Jao, Yeh, & Cheng, 2005). The elastic modulus and Poisson's ratio for the cortical bone of the femur and tibia were 17,000 MPa and 0.3, respectively, (Chatarapanich *et al*., 2009). The talus bone and foot bone structure were assigned the properties of 7,300 MPa for the elastic modulus, and 0.3 for Poisson's ratio (Cheung, Zhang, Leung, & Fan, 2005). All cartilages were assigned the elastic modulus of 12 MPa and Poisson's ratio of 0.3 (Chantarapanich *et al*., 2009). For foot skin, the elastic modulus was assigned according to the model by the work of Agache, Monneur, Leveque, and De Rigal (1980).

2.4 Contact and boundary conditions

In the analysis, all contact bodies were deformable. The analyses were under frictionless modes to simplify the calculation for such large FE models. Deformable joint contacts of the body were attached to each other to simplify the analysis. The displacement was fully fixed for all degrees of freedom at the floor surface. A weight of 540 N of vertical load was distributed on 84 nodes and applied at the lumbar facet of L1. This loading boundary condition technique applied to the bone structure in this study was taken from the work of Filardi (2015). The FE models for the biomechanical analysis without insole and with insole including their boundary conditions are shown in Figure 1.

3. Results

3.1 Stress analysis

Table 1-3 shows the maximum stresses found at the focused areas of the body, which included stress at the intervertebral discs, hip cartilages, knee cartilages, talus cartilages, foot bones, and foot skins.

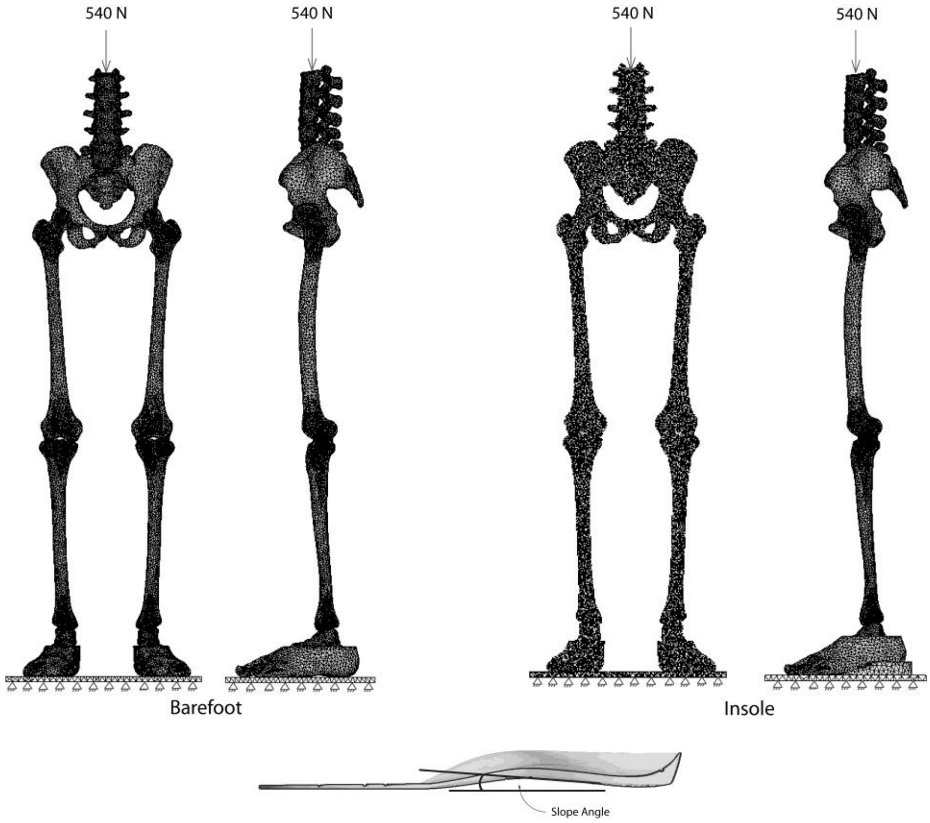

Figure 1. FE models.

Table 1. Stress at the intervertebral discs and hips (kPa).

Case	Region						
	Intervertebral Disc					Hip	
	L1-L2	L2-L3	L3-L4	L4-L5	L5-S1	LT	RT
Barefoot	341.1	75.0	113.6	82.1	92.7	126.1	215.8
0 degrees	341.1	75.0	113.6	82.2	89.5	112.1	151.4
5 degrees	341.1	74.9	113.5	82.2	89.9	119.3	162.4
10 degrees	341.1	75.0	114.0	83.4	91.6	112.2	155.9

Table 2. Stress at knees (kPa).

Case	LT			RT		
	Medial	Lateral	% Difference	Medial	Lateral	% Difference
Barefoot	96.4	102.3	+5.7	98.1	90.8	+7.5
0 degrees	103.1	109.2	+6.1	93.0	97.2	-4.5
5 degrees	107.5	98.2	-9.5	105.7	98.9	+6.4
10 degrees	99.8	99.6	-0.2	97.5	97.3	-0.2

Table 3. Stress at foot region (kPa).

Case	LT			RT		
	Ankle	Foot Bone (Heel)	Foot Bone (Metatarsal)	Ankle	Foot Bone (Heel)	Foot Bone (Metatarsal)
Barefoot	130.9	513.7	518.4	126.7	555.2	548.3
0 degrees	230.8	812.6	260.3	178.5	794.9	321.8
5 degrees	222.6	682.1	231.9	202.7	872.3	321.4
10 degrees	215.4	402.4	248.0	204.4	523.8	272.6

3.1.1 Stress at intervertebral disc

Stress at intervertebral disc level L5-S1 in all insole cases was lower than the barefoot case. In addition, a higher slope tended to produce more stress than the lower slope. No differences in stress magnitudes were observed among any of the insole slopes and barefoot at intervertebral disc levels L1-L2, L2-L3, and L3-L4. Stress at intervertebral disc level L4-L5 was higher as the slope increased.

3.1.2 Stress at hip articular cartilage

Stress at the hip articular cartilages for cases of any insole slope was lower than the barefoot. No difference in stress values was observed at different insole slopes for either left or right hip joints.

3.1.3 Stress at knee articular cartilage

Stress at the knee articular cartilages were monitored in both medial and lateral compartments. The magnitude of stress at the cartilage in the barefoot case was not different than the insole cases. A difference in stress between the medial and lateral compartments was computed according to Chantarapanich *et al*. (2009). The results showed that the stress values in the barefoot case ranged from +5.7% to +7.5%, whereas the values for the insole cases ranged from +6.1% to –4.5%. The positive sign indicates that stress on the medial compartment was higher than on the lateral compartment and the negative sign indicates the reverse meaning. At a 10-degree slope, there was uniform distribution between the medial and lateral sides on both knees and the values were –0.2%.

3.1.4 Stress at ankle cartilage

Interestingly, stress on the ankle cartilages for the insole cases were found to be higher than barefoot and ranged from 64.6% to 76.3% for the left ankle and from 29.0% to 61.3% for the right ankle. There were slight differences for the ankle stresses at various slopes.

3.1.5 Stress at foot bone

Foot bone stresses were concentrated around the talo-navicular joint and metatarsal regions (Figure 2). The insoles with 0- and 5-degree slopes revealed higher stress levels than barefoot at the talo-navicular joint region. Only the insole with a 10-degree slope presented a lower stress than barefoot. The magnitude of stress at the metatarsal regions in all insole slope cases was lower than barefoot.

3.2 Foot pressure

Barefoot pressures were high around the forefoot and heel regions (Figure 3). After insertion of an insole below the foot, pressure was distributed to other regions. However, stress from the 0-degree slope angle was still high around the ball of the foot and heel regions. Stress was distributed almost uniformly around the mid-foot for the insole with a 5-degree slope. High pressure regions were still observed around the heel, but the regions shrunk to a smaller area compared to barefoot. The insole with a 10-degree slope exhibited pressure around the mid-foot; however, a relatively low pressure was observed in other regions.

4. Discussion

This study utilized the FE method to analyze the biomechanics of a large domain of bones and joints. The FE model covered the lumbar spine and downward towards the feet. This domain included most of the significant body parts which allowed monitoring of the biomechanics of the bones and cartilages affected by an insole. This was unlike most of the previous biomechanical studies related to insoles where the analyses covered the body domain only from just above

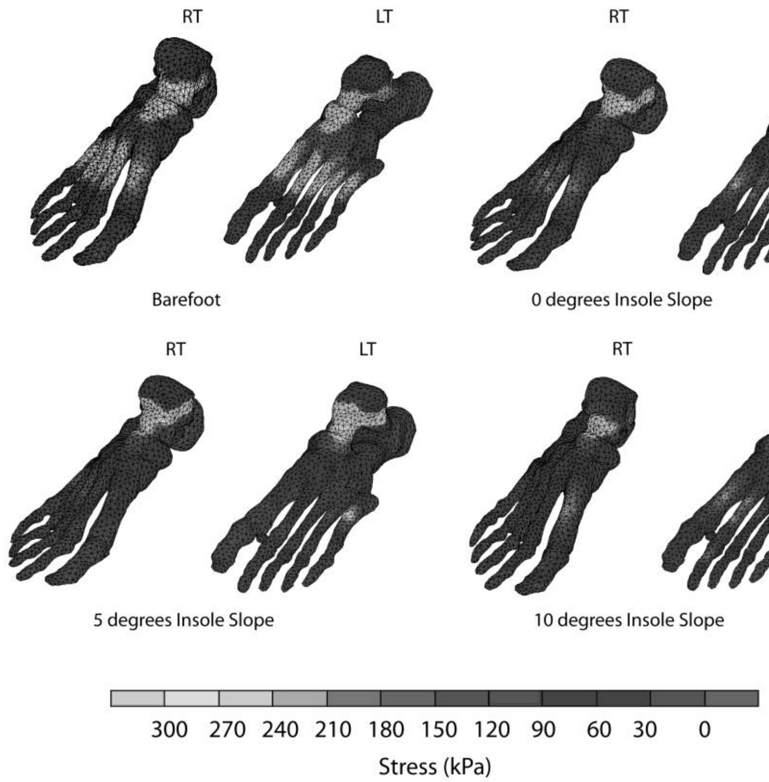

Figure 2. Foot bone stress.

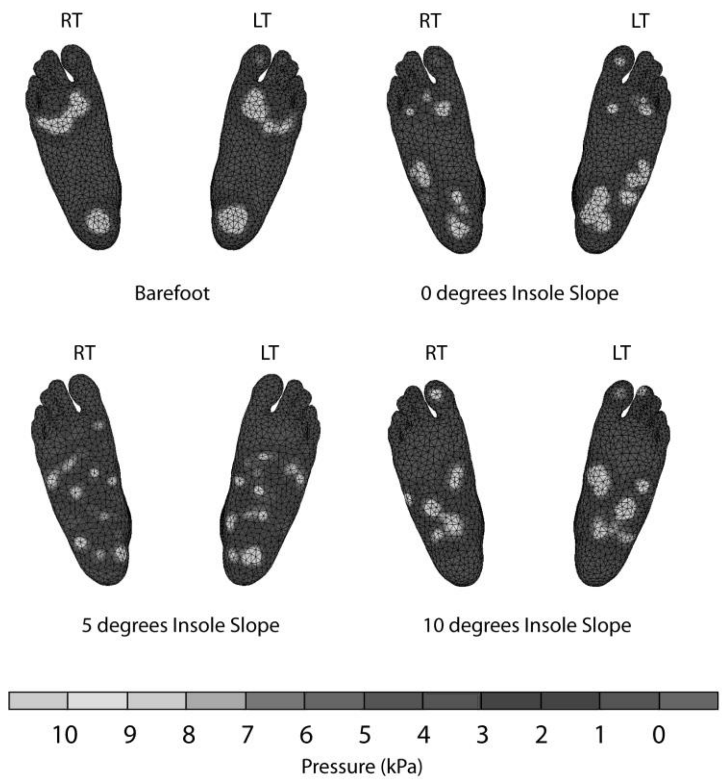

Figure 3. Foot pressure.

the ankle joint and feet (Taha, Norman, Omar, & Suwarganda, 2016; Chen, Lee, & Lee, 2015; Guiotto, Sawacha, Guarneri, Avogaro, & Cobelli, 2014). Only the work of Liu and Zhang (2013) included the domain up to the knee joint in their analysis. The results from this study can contribute beneficially to insole design.

No significant stress reductions were found at intervertebral disc levels L1-L2, L2-L3, L3-L4, and L4-L5 in any of the insole cases compared to barefoot. This finding correlated in some degree to the previous clinical studies of Chuter, Spink, Searle, and Ho (2014) and Mattila *et al*. (2011) who reported that insole or orthotic shoes were not associated with low back pain (LBP) prevention. One of main causes of LBP observed from clinical diagnoses, even with the use of an insole, is still lumbar disc disease (Mattila *et al*., 2011). No specific portions of lumbar disc disease were mentioned in the report. However, stress reduction was found only at intervertebral disc level L5-S1 which was observed by the authors and was correlated with the use of an insole. The slope of the insole influenced the stress only at intervertebral disc level L4-L5 and a higher slope tended to raise the stress magnitude. Thus, insole design should not change the foot alignment in such a way that the heel is the same level or lower than the mid-foot.

An insole also reduces the stress exhibited on the hip joints. This signifies an advantage of insole performance in the prevention of hip osteoarthritis. However, stress at the knee joint for the insole cases was reduced slightly to a lower magnitude compared to barefoot. In addition, no significant shift of maximum stress from one side to the contra-lateral side of cartilage was observed. Shifting stress from a degenerative cartilage to healthier cartilage is preferred to correct the mechanical axis of either a varus or valgus deformity into neutral alignment. To correct the alignment, an insole needs to have a wedge shape form in the medial-lateral direction (Liu & Zhang, 2013). An insole without a wedge slope form, that is similar to the insole in this study, is then not possible. Thus, an insole without a wedge shape is more proper for patients without malalignment of the mechanical axis.

An insole raised the stress at the ankle around the regions of the talo-navicular joint and metatarsal bones. The magnitude of stress at the regions associated with a lower insole slope presented with lower stress. Low insole slope lifted the heel higher than the forefoot which is analogous to a high heel shoe that shifts the center of pressure (CoP) to the forefoot (Sun *et al*., 2017). The moment caused by bodyweight then rises corresponding to an increased moment arm as the CoP shifts towards to forefoot. An insole should not be designed in such a way that the heel is higher than the forefoot. As a result, 5- and 10-degree insole slopes are then more appropriate than a 0–degree slope.

Without an insole, there is no support under the foot. The bodyweight transfers to the floor through the foot contact regions which are the heel and forefoot. Pressure is concentrated around these regions. An insole is a support structure which conforms to the surface of the foot. Application of an insole increases the contact area compared to the barefoot (Raspovic, Newcombe, Lloyd, & Dalton, 2000). High pressures at the forefoot and heel regions are reduced to a lower magnitude. Concentrated foot pressure around the forefoot and heel regions, which is observed in barefoot standing, changes to a more distributed pattern after the insoles are placed under the feet. This is also due to the elasticity of insole materials which influence the reduction of foot pressure and redistribute the foot pressure (Ibrahim, El-Hilaly, Taher, & Morsy, 2013; Tong, & Ng, 2010). The foot pressure pattern found by this study agreed with clinical studies (El-Hilaly, Elshazly, & Amer, 2013; Kato *et al*., 1996).

A different slope of insole corresponds to a different pressure distribution. The pressure produced from the application of a 5-degree insole slope was distributed uniformly around the mid-foot area. For the 0–degree slope, pressure was still concentrated around the heel and forefoot, even though the high pressure area was reduced and redistributed to the mid-foot regions. In addition, although pressure produced from a 10-degree insole slope was low in magnitude around the heel and forefoot regions, the pressure was still concentrated at the mid-foot region and shifted medialwards. Thus, a 5-degree insole slope is considered to be more optimal than a 10-degree slope.

Good candidates of insole slope are 5 and 10 degrees. A 5-degree insole slope presented better reduction in stress at intervertebral disc levels L3-L4, L4-L5, and L5-S1 than the 10-degree insole slope. Greater stress in the knee and ankle joints was observed in the 5-degree insole slope compared to the 10-degree insole slope. Application of the 5-degree insole slope may reduce the risk associated with LBP better than the 10-degree insole slope; however, it may increase the chance of knee and ankle osteoarthritis. The difference in the magnitude of maximum stress on the knee and ankle joints in both slopes was less than 4%, thus the difference can be neglected. Although, higher magnitude of stress was exhibited on the foot bone structure with the 5-degree insole slope compared to 10 degrees, the stress produced from the 5-degree insole slope may be reduced by use of a metatarsal pad or metatarsal bar (Deshaies *et al*., 2011).

In addition, the anatomy of the lumbar region (Lordotic curve) has a high longitudinal curvature. A previous study (Massey, Donkelaar, Vresilovic, Zavaliangos, & Marcolongo, 2012) revealed intervertebral disc degeneration was associated with high disc stress. In the elderly, the intervertebral disc degeneration usually occurs which subsequently reduces the spinal longitudinal length and may lead to spinal disc herniation. As a result, reducing interverbal disc would reduce the chance of the progression and occurrence of the aforementioned diseases. According to the results, a 5-degree insole slope reduced the stress at intervertebral disc levels L4-L5 and L5-S1 which postulates the reduction of intervertebral disc degeneration. However, this postulation needs to be conducted through a biomechanical analysis in the future.

5. Conclusions

This study analyzed the effects of insole slope to joints and bone covering the low back region and lower extremities by means of FE. In barefoot, high foot pressure was concentrated at the forefoot and heel regions. An insole redistributed the foot pressure to the mid-foot region. An insole reduced stress only at intervertebral disc levels L4-L5 and L5-S1. The other intervertebral discs presented no significant stress reduction. An insole reduced stress at the hip joint but increased stresses at the ankle joint and foot bones. The slope of the insole has a significant effect on the stress exhibited in various joints and bones. Higher insole slopes tend to increase the stress especially in the ankle joint and foot bones. The 5- and 10-degree insole slopes are good candidates since both of them presented similar biomechanical performance. However, the 5-degree insole slope can redistribute the foot pressure better than the 10-degree insole slope. The 5-degree insole is considered to be the more appropriate choice.

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