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Pressure relief and load redistribution by custom-made insoles in diabetic patients with neuropathy and foot deformity

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Abstract

Although custom-made insoles (CMIs) are commonly prescribed to diabetic patients with neuropathy and foot deformity, little quantitative data on their mechanical action exists. The objective was to study the effects of CMIs on in-shoe plantar pressure and load distribution in diabetic patients with neuropathy and foot deformity.

Regional in-shoe peak pressures and force-time integrals were measured in the feet of 20 neuropathic diabetic patients with foot deformity who wore flat insoles and CMIs. Twenty-one feet with elevated barefoot peak pressure or history of ulcer at the first metatarsal head (MTH1) were selected for analysis. Load redistribution between anatomical regions due to CMI action was assessed using a new load transfer algorithm.

In-shoe peak pressures and force-time integrals were significantly reduced by CMIs in the heel and MTH1 regions and increased in the medial midfoot region compared with flat insoles. CMIs were successful in reducing peak pressure and force-time integral at MTH1 in seven of 21 feet, were moderately successful in another seven feet, but failed in the remaining seven. The largest transfer of load occurred from the lateral heel to the medial midfoot region.

CMIs were more effective than flat insoles in offloading MTH1 but there was considerable variability between individuals. The largest offloading occurred in the heel, which is not a region typically at risk for plantar ulceration. The load transfer algorithm is an effective tool to examine CMI action.

Because similar insole modifications apparently exert different effects in different patients, a comprehensive evaluation of custom designs using in-shoe pressure measurement should ideally be conducted before dispensing insoles to diabetic patients with neuropathy and foot deformity.
Introduction

Custom-made insoles (CMIs) are routinely used in clinical foot care settings for diabetic patients with neuropathy and foot deformity. The success of such insoles and the associated footwear is primarily evaluated based on whether the patient remains free of ulceration while wearing the footwear and insoles. Elevated barefoot plantar pressure has long been associated with plantar ulceration.\textsuperscript{2,14} Therefore, one of the major goals of any footwear intervention must be to protect the foot at sites that are at risk for plantar ulceration or re-ulceration by reducing pressure to a level below some (currently unknown) threshold for ulceration.

CMIs are prescribed primarily, if not exclusively, to protect the plantar surface of the foot. However, little quantitative data is available on if and how CMIs achieve their action. Among the putative methods that have been discussed is the redistribution of load to adjacent foot regions through accommodative molding - the ‘total contact’ concept - and the incorporation of additional ‘reliefs’ or, conversely, elevated contact surfaces such as medial arch supports and metatarsal pads, which go beyond merely providing a mirror image of the plantar contour.\textsuperscript{4,7,9}

Several authors have reported successful relief of dynamic plantar pressures and load at the metatarsal heads (MTHs), a plantar region where ulcers commonly occur.\textsuperscript{1,5,10-12} However, others have found no significant offloading effects of molded insole interventions.\textsuperscript{2,8,13} The discrepant results from these various studies are likely a function of different approaches to insole manufacture, subject selection, and experimental procedure.

The lack of a standard and comprehensive method of pressure and load analysis in these studies has prevented a better understanding of the mechanisms by which insole modifications act to relieve pressure and redistribute load under the foot. One of the methods by which this relief can be achieved is through comparison of the load distribution patterns from a CMI and some controlled condition, such as a flat insole, worn sequentially both on the same foot and in the same shoe, from which load transfer analysis can be performed. Therefore, the purpose of the present study was to compare the mechanical behavior at the foot-insole interface of CMIs and flat insoles in diabetic patients with neuropathy and foot deformity and to use this comparison to define a calculation method for assessing load redistribution.

Methods
Subjects

Twenty diabetic patients with neuropathy and foot deformity (13 men, 7 women), recruited from a specialist diabetic foot clinic, participated in the study after giving informed consent. The mean (SD) age, height, and weight of the subjects were 64.4 (11.2) years, 1.73 (0.10) m, and 99.5 (15.7) kg, respectively. Neuropathy was confirmed by a loss of protective sensation on the plantar surface of the foot, as determined by the inability to feel the 10-g Semmes-Weinstein monofilament on the hallux of both feet. All subjects normally wore some form of prescription footwear.

Twelve subjects had prior plantar ulcers (8 unilateral, 4 bilateral). At the time of the experiment, all these subjects had remained healed for at least 3 months while wearing their prescription footwear. Foot deformity, which was a criterion for participation, was assessed subjectively. Among the deformities present were claw toes, hammer toes, hallux abducto valgus, limited joint mobility at the first metatarsal-phalangeal joint, forefoot/rearfoot varus or valgus, prominent MTHs, midfoot Charcot neuro-arthropathy, and amputation of the toes. Subjects also had to be able to walk independently with only minor assistance, in case of balance problems. All procedures were approved by the Institutional Review Board of the Pennsylvania State University, where the work was conducted.

Procedures

Both barefoot and in-shoe plantar pressures were measured during level walking. A Novel EMED-SF pressure platform (Novel USA, Minneapolis, MN), consisting of 1984 capacitance-based sensors, each with an area of 0.5 cm², was sampled at 70 Hz to collect dynamic barefoot plantar pressures during five barefoot left and right foot contacts using a first-step collection method. The platform had been recently calibrated over a range of 0-1300 kPa.

The Novel Pedar system was used to measure in-shoe dynamic pressures. This system comprised 2-mm thick flexible pressure-sensing insoles that were connected by a 10-m long trailing cable to a computer. The pressure-sensing insoles, each consisting of approximately 100 capacitance-based sensors sampling at 50 Hz, were placed between the sock and the insole of the shoe. Directly prior to the experiment, the Pedar insoles used for a particular subject were calibrated over a range of 0-600 kPa, according to the guidelines provided by the manufacturer. Four different Pedar insole sizes were used to accommodate the range of foot sizes in the group. On average, 30 steps for each insole condition were recorded from three trials, with the subject walking within ±10% of a comfortable speed established prior to data collection using photocells placed along a 9-m walkway. No
correction was made to allow for differences in sensor dimensions between in-shoe and barefoot pressure systems.

Two different types of insoles that are frequently prescribed in diabetic foot practice were tested in each subject. The first was a standard 0.95-cm thick flat insole (Figure 1A) made of PPT® (Langer, Inc., Deer Park, NY), a soft, durable, non-moldable, open-cell polyurethane foam. The second was a CMI (Figure 1B) manufactured from open-cell urethane foams of hardness 60-80, assessed using an ASTM type 00 tester. Each CMI was specifically fabricated for this project by a CAD-CAM process in which the barefoot plantar pressure data, footprints and tracings of the subject’s feet were sent to a trained orthopedic shoemaker via the Internet. No designated areas of interest were communicated to the shoemaker. No specific algorithm was used in the design of the CMIs. Rather, the skill and experience of the shoemaker were exploited to produce a CMI that was typical of a device that might be produced in a clinical setting. It is recognized that CMIs are often manufactured based on a negative mould of the foot, a technique not employed here.

The main offloading techniques used for the CMIs were the removal of material under high-pressure areas and the build-up of material at other locations by the provision of what
was effectively a metatarsal pad and a medial longitudinal arch support built into the insole, such as might be accomplished by making a mould of the foot. Substantial ‘heel cups’ were also a feature of the CMIs. A digital representation of a typical CMI is shown in Figure 1C. Once the design was complete, the CMI was fabricated by a numerically controlled milling machine from a homogeneous block of urethane foam. A 0.7-mm top cover and a 2-mm base were subsequently added. Both insoles were tested in the same super-depth shoe (PW Minor & Son, Batavia, NY) with the subject wearing thin seamless nylon socks.

The pressure data were analyzed using Novel-Win, Novel-Ortho and Pedar mobile software (Novel USA). Using the ‘Automask’ and ‘Create a Mask’ programs for the barefoot and in-shoe pressure data, respectively, the foot was divided into 10 anatomical regions: medial and lateral heel, medial and lateral midfoot, first, second and lateral MTHs, hallux, second toe and lateral toes (Figure 2A and 2B, see Appendix, p. 183).

For each region, peak pressure and force-time integral were calculated. The force-time integral is a measure of the force impulse or the load applied to the foot in a given region. (The descriptors ‘load’ and ‘force-time integral’ are used interchangeably throughout this article.) The peak pressure within a region was defined as the maximum pressure recorded by any sensor, even partly loaded, in the region during the stance phase of the walking cycle. The force-time integral for a region was defined as the total sum of pressure multiplied by sensor area and sensor contact time of all sensors encapsulated by the region. Because the force-time integral value depends on the regional surface contact area, a single mask was created and superimposed on all single-step pressure pictures from both flat insole and CMI conditions for a given subject so that a valid comparison between the two insoles could be made.

Twenty-one of the 40 feet in which MTH1 was the region of interest were selected. MTH1 was chosen as the region of interest because it was the most common region for prior ulcers (6/19 ulcers) and/or high barefoot pressures (>700 kPa, 19/40 feet) to occur. These 21 feet (from 14 different subjects) were analyzed as a group and individually. The success of each CMI in offloading MTH1 in comparison with the flat insole was based primarily on changes in peak pressure - because this parameter has been associated with plantar ulceration - and secondarily on changes in force-time integral. The CMI was considered successful when both peak pressure and force-time integral at MTH1 were significantly reduced, moderately successful when only peak pressure was significantly reduced without change in force-time integral, and a failure when no significant reduction occurred in peak pressure, irrespective of the effect on force-time integral.
Although one of the clinical objectives of therapeutic footwear is to reduce peak pressure at a given site, the mechanical strategy to achieve this reduction is, most frequently, transfer of load from one region to another. For example, a CMI with a very high medial arch support may transfer load from the forefoot to the midfoot compared with a flat insole. Although the regional impulse or load may change, the total impulse on different insoles does not change if the gait remains the same. Thus, the integral of the force-time curves in the same anatomical regions in two different insoles can be directly compared to determine the transfer of load from one region to another that has been achieved by a given insole.

Inter-regional load transfer or redistribution was assessed quantitatively in the 21 selected feet using a new load transfer algorithm (LTA). The basic principle used in the calculations is that transfer of load takes place from a region where load is reduced by the CMI to a region where load is increased in comparison to loads of the flat insole. However, certain 'rules' for this calculation are required, since the 10 regions are an under-determined system for which more than one solution is possible. Details of the algorithm used are presented in Appendix A (p. 139).

Statistical analysis

The data were analyzed statistically with univariate analysis of variance using SPSS (SPSS, Chicago, IL). For the analysis of the group average data, ‘insole condition’ was the fixed factor and ‘subject’ the random factor in the model. Bonferroni adjustments (significance level of 0.05 divided by number of comparisons) were used for the multiple comparisons made. For the analysis of the individual data, ‘insole condition’ was the fixed factor and ‘trial’ was the random factor in the model (P < 0.05). Pearson correlation coefficients were calculated between selected variables of interest (P < 0.05).

Results

The mean (SD) walking speed with both flat insoles and CMIs was 0.83 (0.31) m/s. The mean absolute intra-subject difference in walking speed between the two insole conditions was 1.6% (range 0 – 4.7%).

Mean peak pressure was significantly lower in the CMIs than in the flat insoles in the medial and lateral heel and MTH1 region. In the medial midfoot and lateral toes, peak pressure was significantly higher with CMIs. In the other foot regions, no significant differences between the conditions existed (Table 1). Total force-time integral was slightly, but not significantly, lower for the CMIs than for the flat insoles. Force-time integral in the
CMIs was significantly lower in the lateral heel, MTH1, and lateral MTHs. In the medial midfoot, force-time integral was substantially larger by 154% in the CMIs ($P < 0.05$). No differences in force-time integral were present between the insoles for the other foot regions (Table 1).

### Table 1. Mean (SD) values of barefoot and in-shoe peak pressures and in-shoe force-time integrals for 21 feet in which MTH1 was the region of interest

<table>
<thead>
<tr>
<th>Region</th>
<th>Peak pressure (kPa)</th>
<th>Force-time integral (N·s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Barefoot</td>
<td>CMI</td>
</tr>
<tr>
<td>Medial heel</td>
<td>414 (222)</td>
<td>189 (45)</td>
</tr>
<tr>
<td>Lateral heel</td>
<td>318 (105)</td>
<td>188 (42)</td>
</tr>
<tr>
<td>Medial midfoot</td>
<td>67 (51)</td>
<td>118 (23)</td>
</tr>
<tr>
<td>Lateral midfoot</td>
<td>174 (182)</td>
<td>121 (25)</td>
</tr>
<tr>
<td>MTH1</td>
<td>911 (217)</td>
<td>255 (81)</td>
</tr>
<tr>
<td>MTH2</td>
<td>534 (268)</td>
<td>183 (35)</td>
</tr>
<tr>
<td>Lateral MTHs</td>
<td>465 (241)</td>
<td>153 (28)</td>
</tr>
<tr>
<td>Hallux</td>
<td>485 (383)</td>
<td>201 (83)</td>
</tr>
<tr>
<td>Toe2</td>
<td>156 (161)</td>
<td>130 (54)</td>
</tr>
<tr>
<td>Lateral toes</td>
<td>140 (103)</td>
<td>120 (47)</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

* Percentage of total force-time integral

$^a$ Significantly different from CMI ($P < 0.05$)

### Table 2. Distribution of 21 feet with MTH1 as region of interest into 'success' categories.

<table>
<thead>
<tr>
<th>Peak Pressure</th>
<th>Force-time Integral</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Increased</td>
</tr>
<tr>
<td>Increased</td>
<td>0</td>
</tr>
<tr>
<td>Decreased</td>
<td>0</td>
</tr>
<tr>
<td>Unchanged</td>
<td>0</td>
</tr>
</tbody>
</table>

$^a$ Successful, $^b$ Moderately successful, $^c$ Failures.

'Success' was based on the statistical effect of the CMIs on MTH1 peak pressure and force-time integral when compared with the flat insoles.

The CMIs were successful in seven feet, moderately successful in another seven, and failed in the remaining seven in their effects on peak pressure and force-time integral at MTH1.
when compared with flat insoles (Table 2; see Methods for definitions of success). The correlation coefficient between change in peak pressure and force-time integral at MTH1 was 0.06 ($P = 0.8$).

All 21 feet examined showed an increase in medial midfoot force-time integral (range 10.7 - 82.6 N·s) and a decrease in lateral heel force-time integral (range 8.6 - 58.4 N·s) when compared with measurements from the flat insoles, and these changes were significantly correlated ($r = -0.80$, $P < 0.001$). The correlation coefficient between load changes in the medial midfoot and MTH1 was not significant ($r = -0.31$, $P = 0.17$).

The LTA revealed that the largest mean load transfer achieved by the CMIs occurred between the lateral heel and medial midfoot (Figure 3). This was 3.3% of the total load and 15.6% of the lateral heel load applied to the flat insoles. Load transfer between MTH1 and medial midfoot was on average 7.8 N·s and amounted to 1.2% of the total load and 8.6% of the MTH1 regional load. The total amount of load (mean (SD)) transferred due to the CMI action was 59.4 (23.7) N·s or 9.4% of the total applied load in the flat insole condition.
A case in which the CMI was highly successful in achieving pressure relief and load redistribution is shown in Figure 4A. The patient was a 60-year-old man (1.87 m, 119 kg). He had a long history of neuropathic ulcers at multiple sites on both feet. He had experienced prior ulcers at MTH1 on the right foot. Acquired foot deformities included a clawed fourth toe, hallux abducto valgus, and limited joint mobility. Barefoot peak pressure at MTH1 was 1058 kPa. In-shoe peak pressure at this site was significantly lower with the CMI than with the flat insole (mean 278 kPa vs. 441 kPa) and force-time integral was also significantly lower, by 24.6 N·s or 25%. Nearly all load transferred from MTH1 was directed towards the medial midfoot. This load equaled 3.8% and 22.7% of the total load and regional MTH1 load, respectively. Moreover, this transfer contributed most to the load increase in the midfoot region (57% of 40.6 N·s).

![Figure 4](image)

**Figure 4.** Two case studies in which the CMI was successful (A) and not successful (B) in transferring load from MTH1 to other foot regions. The regional differences in normalized force-time integral (in N·s) between the CMI and flat insole are shown in (A) and (C) and the resulting inter-regional load transfer values in (B) and (D), respectively. The breadth of the arrows is proportional to the absolute amount of load transfer.

An example of a failure in the mechanical action of the CMI is shown in Figure 4B. This profile is from a 64-year-old man (1.82 m, 90 kg) with a history of ulceration on the second toe of the left foot and several acquired deformities (limited joint mobility, clawed third toe, prominent first to fifth MTHs, and hallux abducto valgus). His barefoot peak pressure at MTH1 was 957 kPa. In-shoe peak pressure was 231 kPa in the flat insole and 216 kPa in the CMI, which was not significantly different. The force-time integral at MTH1 was larger

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by 0.5 N-s in the CMI (not significant). A net load transfer of only 0.1 N-s occurred between MTH1 and the medial midfoot, whereas much larger transfers occurred from the lateral heel and lateral MTHs to the medial midfoot.

**Discussion**

This study has shown that, on average, this type of CMI, intended for at-risk neuropathic feet, can reduce peak pressure and force-time integral at MTH1 by 16% and 8%, respectively, compared with a thick, flat, over-the-counter cushioned insole. It remains to be demonstrated whether or not such alterations are sufficient to reduce tissue loading at this site below the threshold for injury. Neither is the relative importance of reductions in peak pressure or force-time integral for prevention of tissue injury established. These results show a lower therapeutic effect than that reported by Lord and Hosein\(^\text{10}\), who found a reduction of 34% in peak pressure under MTH1 in six diabetic patients wearing molded inserts. Novick et al.\(^\text{11}\) found a 78% reduction in peak pressure at this site in healthy subjects wearing CMIs. Others, however, found no significant changes in peak pressure\(^\text{2,5,12}\) and force-time integral\(^\text{12}\) at this location. These discrepant results are likely related to the use of different insoles, subjects, and experimental procedures, which makes these studies difficult to compare.

Although use of CMIs produced a significant overall reduction in peak pressure and force-time integral at MTH1, an important finding of this study was that 7 of 21 insoles designed by an experienced orthopedic shoemaker were not successful in offloading the foot at MTH1 compared with a simple over-the-counter flat insole. It should be recalled that, in an effort to model the delivery of footwear to a remote location, the orthopedic shoemaker did not actually examine the subjects in the study, but built the insoles based on an examination of barefoot plantar pressures, footprints, and foot outline data. A subjective analysis of the demographic data, the type of acquired foot deformity, and MTH1 barefoot peak pressure, together with observations of the shape and structure of the CMIs, could not discriminate between successful cases and failures. It is possible that more information (such as three-dimensional foot shape, internal foot architecture, and gait characteristics) and/or a more systematic analysis may be required for finding the individual determinants of successful offloading and thus for prescription footwear to be effective. The above results suggest that the effectiveness of a CMI should be measured, where possible, with in-shoe pressure devices to ensure efficacy before it is dispensed to the patient.

A low and non-significant correlation was found between change in peak pressure and force-time integral at MTH1 (\(r = 0.06\)). Several methodological factors may have played a
role in this outcome. First, peak pressure is obtained from a single sensor in the region, whereas force-time integral is measured over the whole region and, as such, is also dependent on contact area and time. In addition, dividing the foot into 10 major anatomical regions may have obscured significant effects of the CMI within the boundaries of a region, for example, in MTH1, where intra-regional pressure redistribution would have reduced peak pressure but kept force-time integral unchanged. Distal parts of the medial arch support may have been included in the MTH1 mask, which likely affected regional force-time integral to a greater extent than peak pressure.

In each of the 21 analyzed feet, the medial arch support proved to be highly effective in transferring load from adjacent regions to the medial midfoot. It was responsible for a 31% increase in peak pressure and a 144% increase in force-time integral in the medial midfoot when compared with the flat insole. This region accounted for 4% of the total force-time integral with flat insoles, but for 11% with CMIs (Table 1). The LTA showed that the four largest inter-regional load transfers occurring within the foot were directed towards the medial midfoot (Figure 3). It is likely that increases in contact area and contact time in this region contributed to the large increase in load and, presumably, explained why this increase was larger than the increase in peak pressure. Although neuropathic plantar ulcers due to repetitive stress are rarely found in the midfoot, avoiding a large increase in peak pressure at the medial midfoot is important so as to avoid any localized damage to the soft tissue of the medial arch, which is not well adapted for weight bearing. Brown et al.\textsuperscript{5} also found increased midfoot peak pressures in 10 healthy subjects wearing CMIs or arch supports inside an extra-depth shoe. Novick et al.\textsuperscript{11} found midfoot peak pressure to be increased by 114% in CMIs when compared with flat insoles in 10 healthy participants. These results suggest that a medial arch support should be a consistent feature in the design and fabrication of CMIs for neuropathic diabetic patients with foot deformity.

The CMIs were also very effective in the heel region. Peak pressure decreased substantially in both medial and lateral heel regions in comparison to the flat insoles, and force-time integral decreased significantly in the lateral heel (Table 1). In the medial heel, pressure may have been redistributed within the boundaries of the region by the arch support, which almost certainly extended into the medial heel mask in some cases. Thus, the medial heel force-time integral was unchanged. The decrease in peak pressure and force-time integral in the heel is most likely caused by two mechanisms: pressure redistribution through the effect of the medial arch support and bilateral cupping of the heel. The LTA calculations clearly demonstrate that load was transferred away from the lateral heel to the medial midfoot (Figure 3). Heel cupping is established by molding the insole around and up the periphery of the heel. The soft tissue of the heel pad is presumably maintained by the CMI in position underneath the bony prominences of the calcaneus, whereas it is displaced in a flat insole.
Brown et al.\textsuperscript{5}, in healthy subjects wearing CMIs, and Albert and Rinoie\textsuperscript{1}, in diabetic patients using custom-made medial arch orthotics, also showed significantly reduced heel peak pressures. Novick et al.\textsuperscript{11} and Ashry et al.\textsuperscript{2}, however, did not find significant effects of CMIs on peak pressure in the heel. Potential differences in size of the medial arch support and/or degree of heel cupping may explain these opposing results.

Most investigations that study the mechanical behavior of CMIs or custom orthoses simply draw conclusions on the pressure-redistributing effect of these interventions based on changes in peak pressure in one or two adjacent regions in the foot. Postema et al.\textsuperscript{12} raised the complexity of the analysis by examining changes in force-time integral in selected regions of the forefoot. However, when force-time integral is measured over the entire surface of the foot, the principle that load decrease in one region automatically results in load increase in another can be used to determine, by altered load distribution, the mechanism by which CMIs work. The LTA was developed with this principle in mind.

As the analysis shows, the largest transfer of load was established between the heel and midfoot regions. A high and significant correlation between load loss and gain in these regions ($r = -0.80$) confirms this association. Presumably, this strong relationship also explains the low and non-significant correlation found between load decrease in MTH1 and load increase in the medial midfoot ($r = -0.31$); the large change in midfoot loading is a result of offloading the heel, not the forefoot. Thus, the dominant effect of the CMIs was expressed in regions that are less at risk for plantar ulceration. The lower absolute values and large inter-subject variability in MTH1 to medial midfoot load transfer (Figures 3 and 4) demonstrates the inconsistency in successful offloading of MTH1 by the CMIs used in this experiment. It should be kept in mind that the load transfer patterns reported here apply specifically to the conditions and subjects tested in this study and are examples of the range of possibilities. A different set of patterns could have been obtained had a different orthotist produced insoles for a different set of feet.

\textit{Conclusions}

This study provides a perspective on alterations in the loading of the feet in a group of patients with diabetic neuropathy and foot deformity who were using CMIs. On average, the CMIs significantly reduced peak pressures and force-time integrals in MTH1 (region of interest) when compared with flat insoles, but their mechanical effects were much larger in more proximal regions of the foot, which are less at risk for plantar ulceration. In particular, dramatic pressure reductions were achieved in the heel as a result of load redistribution by the highly effective action of the medial arch support and, presumably, by cupping of the heel. These effects were very consistent across subjects. Despite significant group results at
MTH1, the CMIs were variable in their pressure-relieving and load-redistributing effect on an individual level, and no improvement compared with flat, over-the-counter insoles was achieved at this target site in seven of the 21 analyzed feet.

These results suggest, as have several prior studies, that based on present knowledge, whether derived from information on barefoot plantar pressure and foot outline alone or from a negative or positive cast from the patient’s foot, experts cannot predictably make efficacious customized devices. Although this statement is based only on measured offloading, high re-ulceration rates in specialty clinics are consistent with this statement as well. We therefore conclude that the effects of CMIs must be thoroughly and systematically examined before we can confidently prescribe such insoles to neuropathic diabetic patients with foot deformity. We suggest that clinicians need to evaluate measurements of in-shoe plantar pressures, including load redistribution patterns (obtained by using a method such as the LTA), to prescribe effective CMIs.

Acknowledgements

The authors thank Mary Becker, Registered Nurse, for her contributions in subject recruitment and screening, data collection, and data analysis. This work was supported by the National Institutes of Health grant HD037433 and by ConvaTec, Inc. Jan Ulbrecht and Peter Cavanagh are principles in DIApedia LLC (State College, PA, USA), a company that performs research and development in the area of diabetic foot disease.
Appendix A. Load transfer algorithm

The force-time integral values in the flat insoles were first normalized to those in the CMIs to account for small discrepancies (average 3.8%, maximum 7.4%) in total foot force-time integral between the flat insoles and CMIs. Normalization was achieved by multiplying all regional force-time integral values in the flat insoles by the ratio of the total force-time integral in the CMIs and flat insoles.

Because there is not a unique solution to the calculation of load transfer between anatomical regions in the foot when comparing a CMI with a flat insole, a set of ‘rules’ for the calculation were formulated:

Rule 1. The foot is divided into 3 levels: heel (level 1), midfoot (level 2), and forefoot (MTHs and toes, level 3). Load transfer calculation starts in the heel, followed by the forefoot regions in order of decreasing load gain and, finally, by the midfoot.

Rule 2. Load transfer can occur between adjacent anatomical regions only.

Rule 3. Transfer to one or more adjacent regions of opposite polarity (load gained or lost) is proportional to the amount lost or gained.

Rule 4. Load is evenly distributed over adjacent regions in the neighboring level, in case none of the adjacent regions are of opposite polarity as the principal region.

Rule 5. When the amount of load lost in adjacent regions is not sufficient to completely solve the principal region, this amount is transferred and calculation should continue with the next region.

Rule 6. To balance the solution at the end of the calculation process, regions with residual loads are solved (in order of decreasing load gain) by transferring load along the shortest route from non-adjacent regions.

The application of these LTA rules is presented in Figure 5A-D for one complex example in the study, chosen because all rules had to be applied to solve the problem. First, the heel is solved (Rule 1). The only adjacent region of opposite polarity to the two heel regions is the medial midfoot (Rule 2, Figure 5A). The total load that needs to be transferred from the heel (20.7 + 15.1 = 35.8 N-s) exceeds the load gain in the medial midfoot (26.2 N-s). By proportion (Rule 3), 20.7/35.8 (= 58%) of the load in the medial midfoot (= 15.1 N-s) will be transferred from the lateral heel, and the rest from the medial heel (26.2 - 15.1 = 11.1 N-s, solid arrows in Figure 5B). The remaining load in the lateral and medial heel regions is now -5.6 and -4.0 N-s, respectively. Because the midfoot regions now have a polarity that is not opposite to that of the heel regions (i.e., load lost), these remaining loads are evenly distributed over the midfoot regions (Rule 4). Thus, 2.8 N-s from the lateral heel and 2.0
N·s from the medial heel are transferred to both midfoot regions (dashed arrows in Figure 5B).

In the forefoot, the hallux region has the largest load gain and is therefore solved first (Rule 1). Proportional to the amount of load lost in the adjacent MTH1 and MTH2 regions (Rule 3), 66% of the load gained in the hallux is transferred from MTH1 (4.2 N·s) and 34% from MTH2 (2.2 N·s). The remaining loads in MTH1 and MTH2 are now -1.3 and -0.6 N·s, respectively (Figure 5B). The lateral MTHs, the next largest load-gaining region in the forefoot, is solved by proportionate load transfer from MTH2, and the medial and lateral midfoot (Rule 3).

![Figure 5](https://example.com/figure5.png)

Figure 5. The load transfer algorithm (LTA) explained using one complex example in the study. (A) Regional differences in force-time integral between the flat insole and CM. (B) Inter-regional load transfer solving the two heel regions, the hallux and lateral MTHs. The large-font italic style numbers indicate loads remaining after the hallux region was solved. (C) Inter-regional load transfer solving all forefoot and midfoot regions. The large-font italic style numbers are remaining loads before the residuals were solved. (D) Final load transfer diagram in which the breadth of the arrows reflects the amount of load transferred.

After solving this region, the remaining load lost in MTH2 (-0.4 N·s) is not large enough to completely solve the lateral toes region (dashed arrow in Figure 5C). According to Rule 5, after having transferred these 0.4 N·s, we continue with the next load-gaining region, which is Toe2. MTH1 is the only adjacent region with opposite polarity, but again the load lost in
MTH1 (-1.3 N-s) is not large enough to completely solve Toe2 (dashed arrow in Figure 5C).

Both Toe2 and lateral toes regions are solved by transfer of residual loads in the non-adjacent midfoot regions (Rule 6, solid arrows in Figure 5C). This is done in order of decreasing load gain (the lateral toes are therefore considered first) and along the shortest route. The remaining loads in the two midfoot regions are approximately equal (-2.8 and -2.7 N-s). Therefore, half of the 3.4 N-s in the lateral toes is transferred from the lateral midfoot via the lateral MTH region, whereas the other 1.7 N-s is transferred from the medial midfoot, via MTH2 and the lateral MTHs region (0.8 and 0.9 N-s). In the same manner, Toe2 is solved. The final result is a load transfer diagram (Figure 5D) in which the breadth of the arrows reflects the amount of inter-regional load transferred.
References