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Methods for Assessing Stress Under the Pathologic Forefoot

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Pressure analysis of the foot while standing and walking has significantly progressed in the past decade. This is especially true for assessing the pattern and absolute dimensions of force and stress on the sole of the foot. Of considerable practical interest to podiatry, or any field that cares for the foot, is the evaluation of forefoot pain. In this chapter, various pressure analysis modalities are detailed. The author has found that no single current modality provides the ultimate pressure analysis. To assist readers in identifying which modality may be most appropriate for their needs, the benefits and limitations of those available is discussed. To illustrate the role of these devices, several case histories detailing use of the pedobarograph in forefoot pain are presented.

Early investigations provided pressure analysis of the general and nondiscrete patterns of stress affecting the foot. Recent advances have allowed identification of more discrete areas and different types of forces. Thus the first step in identifying foot pressures is to decide what type of stress and force are likely to be found. The force components are classified as vertical, anterior/posterior, shear, and mediolateral shear.

Force transducers have been used for several years to assess foot stress. These transducers are made of deforming materials such as polymers, crystals, and metallic rods. The best known is the force platform, which provides a general impression of the ground reaction forces on the entire foot. Earlier devices had inadequate resolution because of the small number of components assessing the stress under the selected area of the foot. Stresses recorded were low, implying that the results of these instruments could not be relied on or compared. The newer instruments have

more components per area under the foot so that the stress pattern information is more accurate and reliable. There are however, still difficulties comparing results and in defining what is normal.

When evaluating these devices one has to be very critical of the engineering concepts by which they were developed¹:

1. What is the dynamic range of the evaluating instrument? That is, does the device evaluate the range of high and low stresses you expect?
2. Does it evaluate the stress site well?
3. What is the sampling rate?
4. Does it have a high number of sampling cells to resolve the force into its components?
5. Does it respond to the frequency of the applied force?
6. Does it respond directly to the amount of force applied?
7. Is it temperature sensitive?
8. Is it reliable and user friendly?

As clinicians we are interested in evaluating the weight-bearing foot for several reasons. The following historical review will help in understanding the types of devices used in the past and those currently available in both clinical practice and research.

MECHANICAL DEVICES

Devices to record the force or stress beneath the foot have existed since 1882 when Beely had subjects step on plaster-filled sacks, theorizing that the magnitude

of pressure was proportional to the depth of the impression. This method and others like it tended to record the shape of the foot and not necessarily the pedal forces.^{2,4} Another early method of recording stress was based on the deformation of pliable projections protruding from the underside of a mat upon which the subject walks or stands. This stress makes the projections collapse; the area of the mat in contact with the surface beneath increases, and produces a darkened area. The intensity of the inked area is proportional to the applied pressure. A pressure image is produced by an inked mat that leaves a single peak pressure picture on the paper below the sole imprint. The disadvantages of this method are twofold; one is the inability to provide any pressure versus time data, and the other is that the image reaches a maximum intensity after which no further increase in pressure can be detected.^{2,4}

The first mechanical device was Morton's kinesiograph. The projections on this device consisted of longitudinal ridges that pressed an inked ribbon onto a piece of paper and left a series of parallel lines that widened with increasing force.⁵

Elftman² used the principle of collapsing projections, but provided pressure-time data. His device consisted of a black rubber mat with pyramidal projections on the bottom that laid upon a glass plate. A white fluid filled the spaces between the pyramids and provided contrast when the pyramids spread. The image was recorded from below with a 16mm movie camera at 72 frames per second. This deformable projection principle is widely used today in the commercially available Harris mat. It has the advantages of being portable and providing better resolution than the previous devices, and is relatively inexpensive.¹ In 1982, Brand⁶ tried to utilize this device as an insole for patients with insensitve feet caused by Hansen's disease, but this simple method failed to provide quantitative data.

The Sheffield optical pedobarograph developed by Betts and Duckworth⁷ after a proposal by Chodera⁸ is the most advanced modification of the foregoing optical systems. The device consists of a glass plate illuminated at the edges by strip lights and covered at the top by a sheet of opaque white plastic with microscopic projections on the bottom. When unloaded, the light from the sides is reflected internally. When a subject steps onto the plastic, pressing it to the glass,

the internal reflections are dispersed and a light is emitted with an intensity proportional to the pressure. Because the continuous gray scale emitted by the device makes it difficult to recognize subtle variations in pressure, the image is converted to a color contour map by a monochrome television camera, an electronic interface, and a color monitor.⁹

Brand and Ebner⁶ used another simple concept and developed a pressure-sensitive device to provide artificial feedback to patients with insensitve hands and feet (Hansen's disease). The device consisted of socks and gloves made of 2 layers of thin polyurethane foam with microcapsules sandwiched between them. When previously established pressures were exceeded, the microcapsules ruptured, staining the socks. The microcapsules were manufactured to rupture at different pressures. The socks and gloves were used to retrain denervated individuals to apply appropriate pressures on their limbs during daily activities. The microcapsules were not reliable and would rupture, depending on the hardness of the underlying tissue rather than applied pressure. Rupture also occurred with repeated use at low pressures. The devices failed to give any warning if the stresses were in the dangerous level. The investigators suggested the use of some type of warning sensor in conjunction with the microcapsule devices.

ELECTRONIC PRESSURE TRANSDUCERS

The failure to quantify stresses using mechanical techniques has resulted in the development and use of electronic transducers. Although the relative complexity and greater expense of electronic systems has previously confined them mainly to experimental use, increasing reliability and simplicity of operation are increasing their attractiveness as a clinical tool.¹⁰

Each type of electronic device has its own strengths and weaknesses. The suitability for a given application should be evaluated on the basis of performance in several important areas. Cavanagh and Ulbrecht¹ suggested the following guidelines when evaluating a transducer:

1. The presence of the device must not change the gait of the subject nor the quantity of pressure measured.

2. The system should be capable of measuring all magnitudes of pressure encountered in the observed activity.
3. The linearity or relation of input to output must be accounted for. A device that is linear produces a consistent increment of output per unit input over the measured range of values. If the device is non-linear, this must be accounted for when analyzing the data.
4. The apparatus must be capable of recording data at the rate at which the measured phenomenon occurs. If a device with poor frequency response is loaded rapidly the output may be in error.
5. The sampling rate must be adequate.
6. The system should exhibit little hysteresis. The input/output curve should have the same shape loading, as unloading.
7. The spatial resolution of the device should be adequate. If you want the difference in pressure between two 1 cm^2 areas, you cannot find it with a single sensor of 10 cm^2 ;
8. The system must distinguish between pressures that are close in amplitude.
9. Changes in temperature should not affect measured values.
10. There must be an adequate calibration procedure and the system divides it by the known area of that sensor. Therefore pressure values are very sensitive to transducer size. The larger the transducer the smaller the pressure for a given force. This is obvious for discrete sensors, but it must be remembered that matrix devices are also made up of functionally discrete transducers.

Because of the lack of studies comparing the different transducers and the values that result from them, as suggested by Lord³ and Cavanagh and Ulbrecht,¹ "It is unwise to compare pressures read from different types of transducers."

CAPACITANCE TRANSDUCERS

A capacitance transducer consists of two conductive plates or elements separated by a flexible dielectric. As the pressure is applied to the device, the distance between plates decreases; the capacitance then increases, and its resistance to alternating current de-

creases.¹⁰ Capacitance transducers may consist of a single layer of compressible material sandwiched between two conductive layers, or they may contain several capacitors in parallel by stacking several alternating layers of plates and dielectrics. This type of device is inexpensive, stable, and produces fairly linear response, but tends to be thick, which makes it less adaptable for use in shoe transducers.^{1,10}

Schwartz and Heath¹¹ created discrete "piezometric discs" out of layers of paper and bronze that were taped to areas of interest on the soles of the feet. Bauman and Brand¹² reported that they were later discarded due to technical difficulties.

Holden and Muncey¹³ used a single shield-shaped capacitance device about 1/8 inch thick and 2 x 2 in. in area, made of three layers of metal foil and two layers of pimpled rubber. The pressure-sensitive area was approximately 0.5 m^2 . This cumbersome device would fit into any shoe, but it was not demonstrated how it fit into the inside of the shoe. It developed a record by taking a photograph of a cathode ray spot.

Bauman and Brand¹² designed a capacitive system to evaluate areas of high pressure on the footsoles of Hansen's disease patients. They used the transducers designed by the Franklin Institute Laboratory¹⁴ and taped them to regions of interest. The pressure transducer consisted of a 1-mm-thick capacitor with a 1 cm^2 pressure-sensitive area. The device was not accurate at pressures greater than 3.5 kg/cm^2 , so the transducer was connected to a preamplifier. It was unstable with changes in temperature and necessitated zero pressure calibrations using a hydraulic device monitored by a precision strain gauge.

In 1978 Nicol and Hennig developed a flexible matrix of capacitance transducers using a 48 x 24 cm foam-rubber mat with 16 conductive strips on either side. The strips were oriented orthogonally to form 256 transducers, 1 at each intersection of strips. The entire array could be scanned in about 5 ms. The advantage of this system was its adaptability to curved surfaces and the ease with which it could be manufactured in different shapes. The disadvantages to this system, however, were cross-talk between transducers, the compression of the foam rubber, which resulted in nonlinear output with some hysteresis, and the fact that the mat could change the pressure under the foot because of its softness.¹⁵ The nonlinearity problem has since been resolved. The refined device

is sold today as the Novel's EMED system (Novel gmbh 1991).

Coleman¹⁶ (1985) used four Hercules F4-4F capacitance transducers measuring 2 mm in thickness and 9 mm in diameter to quantify the reduction in pressures resulting from footwear modifications. The signals were fed to a thermal chart recorder where the peak pressures were measured with a micrometer.

A larger Hercules F4-4R capacitance sensor was tested by Kothari et al.¹⁷ to determine its applicability for use in shoes. This sensor design could be subjected to high loads without harm because the metal did not yield when the corrugations were flattened out. The device was nearly linear up to 500 kPa.

STRAIN GAUGES

A strain gauge transducer is created by bonding a conductive material to a mechanical member or beam. When the beam is deflected, the conductor is lengthened or compressed, changing its cross section and its resistance.³ The load-measuring devices of this type are many and varied, inexpensive, reliable, and have linear output. The devices may be fragile when smaller and be susceptible to temperature changes. When larger, they are bulky and require firm soft tissue to bend the beam.^{1,10}

Hutton and Drabble¹⁸ designed the first strain gauge device to observe the distribution of loads beneath the foot. Used by Stokes et al. in 1975,¹⁹ the system measured stress medial to lateral aspects of the foot and could be rotated 90 degrees to observe anteroposterior loading.

Ctercteko et al.²⁰ appreciated the uses of the walkway system. Noting the poor resolution provided by the Hutton and Drabble system, they constructed one with better resolution. The Ctercteko device consisted of 128 load cells, capable of providing information on the medial to lateral and anteroposterior axes at the same time.

The first strain gauge used in the shoe was devised by Leriem and Serck-Hanssen²¹ when producing a simple in-shoe device with linear output. The transducer was disc shaped, 2.5 mm thick, and covered by a membrane. As the foot contacted the beam it deformed the strain gauges connected to a Wheatstone

bridge. The 2.5-mm thickness of the transducer was a significant disadvantage.

Soames et al.²² used beam-type transducers similar to those of Leriem and Serck-Hanssen with the exceptions that they were square in shape and only 0.9 mm thick. The sensors were not set into the insole of the shoe, but were placed on the foot. These devices created an indentation in the foot that is inappropriate for the insensitive foot. However, these transducers could be calibrated by the manufacturer.

PIEZOELECTRIC TRANSDUCER

A piezoelectric transducer functions on the principle that certain crystalline structures are piezoelectrically active and function as a bundle of dipoles, with positive charges grouped at one side and negative charges at the other. When mechanical stress is applied to the material, separation of charge occurs proportional to the magnitude and orientation of the stress.³⁻¹⁵ The advantages of this transducer are that smaller loads are produced under the foot and that output is linear and exhibits no hysteresis. Its disadvantages are that it is extremely sensitive to temperature changes. Also, the voltage decays with time, so the device is not suitable for static data collection.

Hennacy and Gunther²³ used commercially available crystals (Vernitron PTZ-54) to build a piezoelectric pressure sensor that was easily calibrated, inexpensive, and capable of recording static and dynamic pressures. Hennig et al.¹⁵ manufactured an insole-shaped piezoelectric transducer of separate 4 mm X 4 mm tiles, in matrix form, to resolve the problems inherent to the capacitance mat that they developed in 1978. This insole form was not used again. Cavanagh et al.²⁴ used this in a tiled form. Hennig and associates used it in a discrete sensor form for several studies involving running and other athletic events.²⁵⁻²⁶ Bhat et al.²⁷ used commercially available piezo film to build a transducer that was thin, inexpensive, and durable. It would not register steady pressures at all, but the linearity of the device was good.

There are many problems inherent in piezoelectric

devices, problems which have discouraged clinical use of piezoelectric transducers.^{1,10}

FORCE-SENSING RESISTORS

Force-sensing resistors (FSR) are made by impregnating an elastic material such as foam rubber with a conductor such as metal or carbon powder. Two conductive sheets sandwich the conductive elastomer. When the sandwich is compressed, the elastic material gives and allows more surface contact of the powdered conductor, which results in decreased resistance proportional to the amount of compression.²⁸

The main advantages of FSRs are their simplicity and very thin cross section. Most investigators who use these devices do so because they are thin and lightweight. The most widely used FSR is sold commercially as a complete pressure recording system called the Electrodynogram (EDG) (Langer Biomechanics Group, Deer Park, NY)²⁹ that was developed in 1979. Early users of the systems encountered difficulties. Misevich³⁰ chose the EDG for its small size and unobtrusiveness but found the early sensors were too thick and too variable to use in any reproducible experiments. In 1984 he used the newer Mylar sensors but cautioned others against using them without first performing rigorous calibration procedures. Brodsky et al.,³¹ using the EDG in 1987, reported it was inaccurate and that its results could not be reproduced. He cautioned investigators to be skeptical until its reliability was improved. Other investigators³²⁻³⁴ used the system with no negative comments but did not discuss if they calibrated the device. It appears to be a good foot timing device.

MAGNETO RESISTOR SENSORS

The magneto resistor uses a semiconductor the resistance of which varies with the strength of the magnetic field in which it is placed. The device was developed by Tappin et al.³⁵ to measure shear forces on the sole of the foot and was used by Pollard et al.³⁶ to examine shear forces in combination with normal stress. A similar but smaller version of the sensor was recently developed by Laing et al.³⁷ The Tappin transducer is constructed using two

stainless steel disks 16 mm in diameter. The upper disk was grooved and attached to the subject's foot. The lower disk had a corresponding ridge which fit into the groove of the upper disk and allowed sliding translation between the two disks along one axis only. A magneto resistor was mounted flush with the floor of the groove, and a magnet was attached to the ridge. When assembled, the magnet and resistor would slide relative to each other. The disks were held together with silicone rubber, which allowed translation of the disks relative to each other and provided a re-centering force. The electrical signal produced was proportional to the movement of the magnet, which was in turn proportional to the applied shear force.³⁵ The transducer is 2.7 mm thick and may be noticeable under the subject's foot; the device developed by Laing and associates is only 1 mm thick.

CLINICAL APPLICATIONS

Clinical devices for the pressure measurement are of two categories, a discrete transducer and the matrix type. If you are interested in evaluating a discrete area of the foot, devices like the electrodynograph, magneto resistors, or piezoelectric devices should be considered. They are small, considerably less expensive, often portable, and easy to use for evaluation of the shoe-foot interface. Placement of the sensors is extremely important. Cavanagh and Ulbrecht¹ stated that the drawback of discrete sensors is that the investigator makes an assumption where the pattern of plantar pressure is before attaching the sensors to the foot. Lord³ warned that discrete sensors might not give accurate measurements because of difficulty in accurately positioning the transducers. Care is required to position the sensors precisely at the intended location, which may be accomplished by radiography,²¹ or by palpation of bony prominences or after location with inked pads.³⁸

One must try to stop any movement of the sensor on the foot surface by applying tape or adhering the sensor. Misevich³⁰ reported that migration of the sensors on the sole of the foot ultimately required them to be in laid into the insole. The discrete transducer can act as a foreign body in the shoe, distorting pressures and affecting the subject's gait.^{39,40} Lake et al.⁴¹

compared plantar pressures between transducers placed on top of the insole and those recessed into it, finding that transducers on top of the insole surface had increased recorded pressures.

When one requires an evaluation of the plantar aspect of the foot, the matrix devices are used. They evaluate the entire plantar aspect giving a relationship of discrete pressures to the entire distribution of plantar pressure. They can be used to determine the cause of a lesion (for example, diabetic) or to compare a pre and postsurgical state. However, the matrices are likely to underestimate the peak pressure as the load may be borne on several sensors at once.²⁵

The matrix devices are usually placed in a walkway (e.g., the EMED-SF, Musgrave FSR, Pedobarograph BTE) or in the shoe (the F-Scan or Micro EMED). The difficulty when the modality is in a walkway or set in the ground is that this only allows for bare foot or outer shoe conditions to be evaluated. Another difficulty is the collection of successive trials, the mid gait collection technique versus the first-step collection technique (which may avoid the targeting problem of midgait collection).^{1,42} The circuitry and subsequent expense of these devices is considerable when compared to the discrete sensor.^{9,15,25}

THE PRACTICAL EVALUATION OF SUBJECTS WITH FOREFOOT DEFORMITY AND PAIN

Four examples utilizing the pedobarograph plantar pressure assessment are presented to show the use of this device in the decision-making process for treatment of the painful forefoot. Significant numbers of surgical forefoot corrections are attempted every year, and many of them are re-surgeries because past procedures have failed. As detailed in the first case study, use of the pedobarograph may help to reduce the number of surgical failures from poor subject selection.

CASE STUDIES

Case Study 1

An 18-year-old Caucasian woman presented with complaints of pain in the great toe joints during the past

Table 35-1. Results of Arthrometric Examination

Anatomic Site	Left	Right
Hip	11° external	frontal plane
Hamstrings	Zero	Zero
Malleolar torsion	18° external	18° external
Ankle dorsiflexion	10°	10°
Subtalar joint	13° inv/8° ev	19° inv/9° ev
Forefoot to rearfoot	6° varus	4° varus
RCSP	4° ev	1° ev
NCSP	10° inv	10° inv
Limb length	Equal	Equal
Tibial influence	9° mv	9° mv

Abbreviations: inv, inversion; ev, eversion; RCSP, resting calcaneal stance position; NCSP, neutral calcaneal stance position.

year. Wearing high-heeled shoes causes aching in the great toe joint. Table 35-1 shows the results of her arthrometric examination.

Visual gait analysis noted a pronated stance phase, with heel contact phase demonstrating rapid heel eversion past the vertical. This heel eversion appeared to continue into late midstance, resulting in an everted heel lift and propelling with a medial rolloff. The hallux valgus functional deficit became very evident in the pedobarograph results for both the left and right foot.

The lack of proper forefoot loading is first noticed from the medial to the lateral column. The first ray demonstrates that the load is inappropriate on the right side until the middle of midstance phase of gait (0.35 s). This type of function is seen in the pronated foot. The hallux valgus deformity creates a lack of forefoot stability at the end of forefoot loading phase. The unstable medial column is related to the forefoot varus, demanding the rear foot continue pronating to enable the forefoot to reach the ground. The compensation of the rearfoot varus also continues the instability of the lateral and the medial columns. The pronation continues about the midtarsal joint until the second through fourth metatarsals are stable in the sagittal plane.

The preparation of the forefoot stability takes three-quarters of the stance period. The demand for stability creates a plantar flexion of the hallux to resist the lateral to medial weight transfer. This occurs in conjunction with the sagittal plane stress that results from

body movement. The hallux stress is larger than the load under the first metatarsal head. The pattern of increased hallux and second metatarsal stress is associated with a decreased first metatarsal stress, defined as a functional hallux limitus (FHL). A functional or structural metatarsus primus elevatus will exhibit pressure findings similar to that described as FHL.

Radiography and clinical evaluation will distinguish between the two conditions. As seen in this case, on x-ray the first metatarsal was elevated; however, it was not short and a tongue-and-groove configuration was noted in the first metatarsal head. The majority of the hallux valgus angle was formed by the distal abductus angulation.

Orthotic functional control of this case should focus on the heel lift period of stance. Control at this time would enable the foot to adequately prepare for the next stance phase. This would be achieved by using an aggressive rearfoot and forefoot varus post, 18-mm heel cup height, and a Morton's forefoot extension. The control and balance of the forefoot loading would be required up to 85 percent of the stance phase (early propulsive phase) of gait.

In this case, what surgical procedure would you contemplate to correct the functional asymmetries in the medial and the lateral columns of the foot? Would a procedure to lower the first ray be sufficient? Would a procedure to elevate the second and third metatarsals also be appropriate to avoid future problems? Could a more aggressive surgical plan such as derotation of the navicular be best? In conjunction with the surgery, would the orthotic therapy be sufficient to support the foot after the procedure?

The answers to these questions become very important when considering the surgical intervention on the functional hallux limitus or metatarsus primus elevatus (MPE) foot type via the first metatarsophalangeal joint arthroplasty with or without implant. One can realize the stress that would develop under the central rays of the foot in the MPE foot type after such a procedure.

Case Study 2

This example is a 61-year-old black man who complained of a painful hallux valgus deformity of his left foot. This subject demonstrates a severe limited ankle dorsiflexion of 90° of both feet with the knees extended. He also presents with a moderate rearfoot

varus and inverted forefoot to rearfoot. Observation of his gait notes an inverted heel at heel strike while at midstance his midtarsal joint is compensated by a fixed pronated position.

Biomechanical function was further assessed while the subject walked over the pedobarograph matrix platform. The forefoot loading is caused by the deviation of the first metatarsophalangeal joint; first metatarsophalangeal joint loading notes practically no stress in the left foot and minimal stress in the right. Severe stress is present under the second metatarsal with a supportive stress by the hallux. The loading characteristic of the hallux and the second ray is seen during the late stages of midstance and the propulsive phase of gait. This represents a hallux limitus and non-supportive function of the first ray. Is this the sign of instability of the first ray or the sign of hypermobility or an adapted elevatus of the first ray? Correlated to the biomechanical findings, a large range of first-ray dorsiflexion and lesser first-ray plantar flexion was noted. The first metatarsophalangeal joint was also restricted in its dorsiflexion (25°).

How much would the orthosis manipulate the patients propulsive function? Would it reduce the need for surgery? Would a plantar-flexory surgery of the first metatarsal assist the balance of the stresses at the forefoot?

Case Study 3

A 32-year-old Caucasian male warehouse worker complained of a painful left foot in the area of the ball of the foot near the third metatarsal head. The tyloma was noted to be concentrated under the site of pain with a fibrous base. The patient described his pain being localized and having a burning/shooting character. He ambulated with a slight antalgic gait; overactivity of the anterior tibial muscle and lesser extensors musculature was also noted. In biomechanical examination, a compensated moderate rearfoot varus and forefoot varus in association with a low axis of the subtalar joint were noted. Ambulation throughout stance was with an everted relaxed calcaneal stance position.

The pedobarograph notes loading of the forefoot that is out of the normal sequence of the fifth to the first metatarsal loading sequence. The loading of the forefoot begins at the time of maximum load (0.20 s)

for the heel, which is about 50 percent of the maximum load to the forefoot. The concentration of the vertical force is under the second through the fourth metatarsal heads. The load under the second and the third metatarsal heads and halluces is at their maximum (at 0.65 s) of the foot during late propulsive phase. The center of pressure line of this patient suggests a moderately pronated function in this foot as well. Similar function is noted in the right foot.

Questions arise as to what therapy would be best to relieve the neuritic symptoms as well as the tyloma in the area of maximum stress. Would a forefoot to rearfoot balance be sufficient in the positive cast or would it be better to use an extrinsic post? Will soft tissue supplementation with accommodation of a 4-mm depth be required? Or would it be best to elevate or shorten the metatarsals along with excision of the entrapped nerve?

Case Study 4

The final example is a patient who had previous metatarsal surgery of the area of interest, the third metatarsal of both feet. The lesions were present for several years before surgical intervention. There was no pre-operative pedobarographs. The lesions were not altered by the surgical intervention. No orthotic therapy was utilized after surgery.

Now the patient presented for a second opinion with deep callused lesions under the fourth metatarsal heads of both feet. In addition she presented biomechanically with compensated forefoot varus of 5 degrees, a rearfoot varus of 4 degrees, and contracted third and fourth metatarsophalangeal joints.

A pedobarograph examination was conducted before and after debridement of these lesions. Debridement of the lesions reduced the abrupt contact of the area of interest. The vertical stress in this area was reduced approximately 70 percent, and the patient's antalgic ambulation improved dramatically.

Therapeutic concerns for these lesions would be further surgical intervention to reduce the contracted metatarsophalangeal joints, an elliptical plantar skin revision with appropriate midtarsal and forefoot orthotic control, or elevation of another metatarsal.

SUMMARY

The aim of this brief review of pressure-evaluating tools is to develop an awareness of the types of mechanical devices available for better evaluation of painful complaints of the forefoot and better patient care. Deformity is generally very obvious, but more depth of thought and insight into function are needed to provide correction. The tools are available now to investigate and to establish the biomechanical parameters of surgical therapy.

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