Description and validation of a novel liquid metal-free device for venous congestion plethysmography

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christ, F., a. Bauer, d. Brügger, m. niklas, I. b. Gartside, and J. Gamble. Description and validation of a novel liquid metal-free device for venous congestion plethysmography. J Appl Physiol 89: 1577–1583, 2000.—We present a newly developed electromechanical sensor with automated calibration for strain-gauge plethysmography (filtrass) and compare it to a conventional mercury-in-Silastic strain-gauge plethysmograph (MSG). Fluid filtration capacity ($K_f$) and isovolumetric venous pressure ($P_{iv}$) of the limb were assessed noninvasively with both devices in 29 healthy volunteers. We found significantly higher $K_f$ and $P_{iv}$ values with MSG [4.6 ± 2.0 × 10⁻³ ml/min⁻¹·mmHg⁻¹·100 ml tissue⁻¹ ($K_f$ units; $K_f U$) and 21.2 ± 8.1 mmHg for $P_{iv}$], than with filtrass, giving values of 3.1 ± 0.8 $K_f U$ and 15.1 ± 7.1 mmHg. Because $K_f$ and $P_{iv}$ are profoundly influenced by the calibration, we investigated the quality of the calibration signal and its impact on the obtained values. We could show that the reproducibility of repeated calibrations was higher with filtrass (58% lower mean ± SD). The data were grouped according to the quality of calibration, and we found no significant difference in $K_f$ and $P_{iv}$ between filtrass (3.0 ± 0.7 $K_f U$ and 15.9 ± 6.9 mmHg, respectively) and MSG with good calibration signal (3.3 ± 0.8 $K_f U$ and 18.6 ± 7.1 mmHg, respectively; no significant difference). However, we obtained significantly higher MSG values (5.6 ± 2.0 $K_f U$ and 23.1 ± 8.4 mmHg, respectively; $P < 0.001$) in the group with a bad calibration signal. We suggest that the filtrass sensor, which performs an automated, standardized calibration procedure and shows a linear signal response to stretch, gives highly reproducible and reliable results and thus is more suitable for routine application.

The strain-gauge technique, first described by Whitney (28), utilizes a mercury-filled elastic tubing as a gauge [mercury-in-Silastic strain gauge plethysmograph (MSG)]. Changes in limb circumference are deduced from alterations in the resistance brought about by changes in diameter and length of the gauge. Although elegant and widely used for the noninvasive measurement of blood flow (2, 28), the metal-filled gauge system has some drawbacks, e.g., the toxicity of mercury, the variable lifetime of gauges, and the potential for producing an inaccurate calibration. The latter may be due to the apparent nonlinearity of the resistance change, most markedly obtained at the extreme ends of the strain-gauge stretch. We developed a new mercury-free strain-gauge design with an integrated automatic calibration device (filtrass 2001, DOMED Medizintechnik, Munich, Germany) that overcomes some of the major drawbacks inherent in the use of liquid-filled sensors for venous congestion plethysmography (VCP). Gamble et al. (11), using a modification of Whitney’s original mercury-filled strain-gauge system (28), described a VCP protocol for determination of fluid filtration capacity ($K_f$) and isovolumetric venous pressure ($P_{iv}$). The same system can also be used for the assessment of venous pressure (5) and factors influencing blood flow (10).

The present paper describes a novel electromechanical sensor, which performs an automated calibration procedure. Most commercially available strain-gauge plethysmographs do not allow for a calibration of the strain-gauge response to stretch. Usually only the resistance of the strain gauge is tested, thus ensuring the integrity of mercury-metal contact within the tubing of the sensor. We used the system described by Gamble et al. (11), which enables a manual calibration. We investigated the quality of the calibration of this mercury-filled gauge while applying two different calibration protocols and compared it to that obtained with the newly developed filtrass system. The studies were performed on a model limb and on human legs. Moreover, we compared the values of $K_f$ and $P_{iv}$ obtained in 29 healthy volunteers using the protocol described by Gamble et al. (11). We were interested in establishing why different investigators have found mean control values for $K_f$ ranging from 2.6 to 5.6 × 10⁻³ ml/min⁻¹·100 ml tissue⁻¹·mmHg⁻¹ ($K_f$ units; $K_f U$) (see Table 1), using a similar VCP protocol and strain-gauge design to that described by Gamble et al. (11).

METHODS

All subjects gave informed consent, and the study was approved by the local ethics committee.

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Table 1. Fluid filtration capacity values found by different investigators using mercury-in-Silastic strain-gauge plethysmography

<table>
<thead>
<tr>
<th>Investigator</th>
<th>Mean ± SE</th>
<th>Median and range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gamble et al. (11)</td>
<td>4.3 ± 0.2</td>
<td>Mean ± SE</td>
</tr>
<tr>
<td>Gamble et al. (7)</td>
<td>3.8 ± 0.4</td>
<td>Mean ± SE</td>
</tr>
<tr>
<td>Gamble et al. (10)</td>
<td>3.3 ± 0.5</td>
<td>Mean ± SE</td>
</tr>
<tr>
<td>Gamble et al. (9)</td>
<td>3.2 ± 0.4</td>
<td>Mean ± SE</td>
</tr>
<tr>
<td>Gamble et al. (8)</td>
<td>4.8 ± 0.4</td>
<td>Mean ± SE</td>
</tr>
<tr>
<td>Jeal et al. (15)</td>
<td>3.8 ± 0.2</td>
<td>Mean ± SE</td>
</tr>
<tr>
<td>Jaap et al. (14)</td>
<td>4.5; 3.2–5.7</td>
<td>Median and range</td>
</tr>
<tr>
<td>Jaap et al. (13)</td>
<td>5.4; 3.5–8.0</td>
<td>Median and range</td>
</tr>
<tr>
<td>Mahy et al. (22)</td>
<td>2.6 ± 0.7</td>
<td>Mean ± SD</td>
</tr>
<tr>
<td>Mahy et al. (21)</td>
<td>3.6 ± 1.1</td>
<td>Mean ± SD</td>
</tr>
<tr>
<td>Mahy et al. (21a)</td>
<td>2.9 ± 0.7</td>
<td>Mean ± SD</td>
</tr>
<tr>
<td>Christ et al. (3)</td>
<td>5.6 ± 1.9</td>
<td>Mean ± SD</td>
</tr>
</tbody>
</table>

Numbers in parentheses are reference numbers.

**Filtrass 2001.** The underlying principle of the filtrass 2001 sensor is fundamentally different from that of the liquid metal-filled Silastic tubes of conventional strain gauges, because no elastic recoil of a distensible component is involved (Fig. 1). Changes in limb circumference are detected with a passive inductive transducer, which has an accuracy of ±5 μm (Fig. 2). An inelastic but flexible plastic line with a diameter of 0.5 mm spanning around the limb connects the transducer to an in-built motor (Fig. 2) with a position accuracy also of ±5 μm and a weight of 12 g. The plastic line glides on a flexible zigzag band with a silicon-coated surface, ensuring low friction and good fixation of the sensor to the limb (Fig. 1). When the limb circumference increases, the passive transducer can be pulled out to a maximum of 4 mm. This is followed by an outward drive of the stepping motor if the length change exceeds 4 mm. This design enables measurement of a maximum change in circumference of 22 mm, which is certainly greater than the circumference changes that we have experienced during our cumulative small pressure step studies (11). With the filtrass sensor, the motor automatically resets to the initial position of the passive transducer before each pressure step. This procedure ensures that the sensor is always operating over the same sensing range, an advantage that is not shared by the Silastic tube system.

The calibration of the sensor is touch free, thus reducing artifacts due to manipulation. The motor applies a pre-tension pull of 1 mm, which is followed by a calibration pull of 4 mm. The response of the inductive transducer to the pull is sampled at 10 Hz and measured in arbitrary units. Deviations from the ideally linear relationship, between motor pull and the response of the passive transducer, are included in the calibration. Three repetitive measurements are automatically performed during each calibration procedure, and the second and the third are compared for the time delay of the response of the passive transducer. The maximum time delay value that is accepted is 500 ms. Moreover, the calibration program will not accept deviations of >200 μm between the measurements. The calibration and data recording procedures are fully automated and computer driven.

Specific pressure step protocols can be written, saved, and selected at the start of a study. Moreover, the protocol can be adapted during a study, thereby enabling allowance to be made for changes in circumstances, e.g., therapeutic interventions.

MSG. MSG has been described in detail elsewhere (11, 28). Briefly, the gauge consists of tubing made of a silicon elastomer, filled with mercury, sealed with molybdenum pins, and balanced against an adjacent temperature compensation coil in a Wheatstone bridge. A double loop is formed around the limb to increase the length of the sensor and, therefore, the accuracy of measurement (28). The loops are positioned 1 cm apart on a calf with measured circumference and kept in place by means of plastic guides on ribbon with a low coefficient of friction. The friction component is decreased further by coating the guides, ribbon, and gauges with silicone lubricant (11). Changes in strain-gauge length cause alterations in electrical resistance of the gauge in the low-voltage circuit.

**Calibration procedures and quality of calibration.** A pretension giving a deflection equivalent to a 5-V vertical movement on the computer screen, i.e., a 50% vertical shift, was found to provide sufficient tension (11). The standard, known increase and decrease in gauge length were then delivered by means of five double turns of the calibration nut. The five up-and-down steps represented the change in signal attributable to known increases and decreases in gauge length, equivalent to ~0.8 V/mm change in length. At the end of the calibration procedure, the input length and voltage signals were translated into a calibration factor, with units of volts per millimeter, which was saved to disk for use during the subsequent off-line analysis.

All MSG data were sampled at 1 Hz for subsequent off-line analysis. Because one objective of this study was to investigate the influence of gauge pre-tensioning protocols on the results obtained, we quantified the quality of the calibration procedure by looking at the slope of the relationship between stretch and signal and by noting the deviation from the ideal linear relationship.

All MSG calibration data were analyzed, off-line, to study the relationship between signal volts displayed on the com-
puter screen and length (micrometers) due to each calibration pull of 5,000 μm. The slope of the relationship between volts and length in micrometers was defined as the calibration factor. A retrospective analysis of 109 previously performed MSG studies revealed a mean calibration factor of 0.83 ± 0.19. We defined a good calibration as lying within ±1.5 SD of this calibration factor. Hence a calibration was considered “good” when the volt-to-millimeter ratio was between 0.55 and 1.11 and “bad” when it lay outside this range. By that means, we excluded the extreme ends of the gauge stretch, in which the nonlinearity is most apparent. Referring to Hooke’s law, a linear mechanical behavior of a coil can be expected only after an adequate pre-tensioning and before it is overstretched. These two situations would result in either a low or a high calibration factor, which were defined as bad calibrations.

**Study design.** In the first study, we tested the reproducibility of the calibration signals of MSG and the filtrass sensors during repeated measurements. Both types of sensors were fixed on a model limb (circumference 420 mm), and the calibration procedure was repeated 11 times. The deviation from the first run was compared for each run and expressed as standard deviation.

**Reproducibility of the calibration in vivo.** We tested the reproducibility of the calibration signal of both MSG and filtrass on seven healthy young (20–24 yr) volunteers (2 women, 5 men). Two MSG and two filtrass sensors were applied in randomized order on both calves (mean circumference 366 (range 320–400) and 364 (range 318–399) mm, respectively). The calibration procedure was repeated 11 times for each of the four sensors. We calculated the deviation of each calibration signal from the first calibration run and determined the standard deviation.

**Comparison of Kf and Piv obtained by filtrass and MSG sensors.** Because the venous congestion protocol, which was common to all studies, has been described in detail elsewhere (11), we will only give an outline of the basic protocol used. Bilateral assessments of 29 healthy subjects, giving a total of 50 analyzable observations, were made (13 women, age 23.3 ± 2.3 yr, weight 58.4 ± 6.1 kg, height 170.3 ± 6.3 cm and 16 men, age 24.0 ± 3.5 yr, weight 76.4 ± 5.3 kg, height 180.1 ± 4.9 cm). The subjects lay supine for at least 20 min before the commencement of the study protocol. The legs were supported on a vacuum mattress with the midcalf points leveled with respect to the right atrium (one-third of the distance from sternal angle to the surface of the supporting bench). Congestion cuffs were fitted bilaterally at mid-thigh level, and the pressure was applied by the in-built filtrass pressure module. This allows a free selection of pressure steps ranging from 1 to 350 mmHg. In the present study, we applied seven cumulative pressure steps of 10 mmHg, which were maintained for 4 min each.

The circumference of each calf was measured before the strain gauge and the filtrass sensor were attached. Close to the maximum circumference of the calf, one MSG and one filtrass sensor were placed on each leg so that a total of four sensors were used. The position in relation to the knee was chosen in randomized order. The MSG sensors were set to the standard tensions and then calibrated at the start of each study, as described above.

**Determination of Kf and Piv.** The determination of Kf and Piv follows the protocol described by Gamble et al. (11) and was used both in MSG and filtrass. Seven cumulative small pressure steps (10 mmHg) were applied without exceeding the diastolic blood pressure. The resulting volume changes were simultaneously measured with the two types of sensors and analyzed off-line. The ambient temperature was continuously recorded throughout the study period using the temperature sensor of filtrass, which is on the surface of the sensor holder.

**Data analysis.** When cuff pressure (Pcuff) is elevated above the existing venous pressure, a rapid increase in limb circumference attributable to altered vascular volume can be seen. A gradual increase in limb volume is observed when Pcuff exceeds filtration pressure at the microvascular interface. This change is caused by fluid filtration. Both the vascular volume and limb volume components can be obtained for each pressure step, because a series of pressure steps are applied. The relationship between Pcuff and limb volume is linear and represents the Kf (11). The intercept on the x-axis is the Piv, the Pcuff that needs to be exceeded to generate net fluid flux. A noninvasive measure of venous pressure at the level of the strain gauge is obtained by determining the intercept of the relationship between the vascular volume component and Pcuff (5). The data obtained with MSG and filtrass systems were blinded and analyzed independently by two investigators.

**Statistical analysis.** For data analysis we used Sigma Stat (SPSS, Chicago, IL). Values are given as means ± SD, and the mean values are compared by the paired Student’s t-test; in cases of nonnormal distribution, we applied the Wilcoxon signed-ranked test. Significance was assumed at P < 0.05.

**RESULTS**

**Reproducibility of the calibration signal (model limb).** The regression analysis of the 11 repetitive measurements with 5 steps on the model limb revealed a $r^2 = 0.9751$ for MSG, which was comparable to $r^2 = 0.9994$ in filtrass. The deviation of 10 repeated manual calibrations from the first calibration of MSG gave a mean ± SD from the first run of ±99 μm compared with ±16 μm with filtrass. The maximum deviation observed in MSG was 395 μm, whereas the automated calibration procedure gave a maximum deviation of 43 μm.

**Repetitive calibrations on human limbs.** The mean intraindividual standard deviation during 10 repetitive calibration runs on human legs was ±132 (range 37–251) μm for the MSG sensor compared with ±56 (range 29–109) μm for the filtrass sensor. The standard deviation during 280 individual calibration procedures in all volunteers ranged from 9 to 327 μm in MSG but was only 13–150 μm for the filtrass sensor. Representative examples of the calibration in vivo are given for MSG in Fig. 3A and for filtrass in 3B. The obvious nonlinearity of the MSG calibration signal and the wider deviation between the different calibration maneuvers can be seen.

**Kf and Piv values for 29 healthy volunteers.** No significant changes in the temperature (either ambient or skin temperature) occurred during the study period. The in-built temperature probe of filtrass, which is located 5 mm above the skin level, representing a mixture of both skin and ambient temperature, was 28.6 ± 1.1°C at the start and 28.3 ± 1.0°C at the end of the study.

Measurement with the MSG plethysmograph gave significantly higher (P < 0.001) Kf values (4.6 ± 2.0 KU) than with filtrass (3.1 ± 0.8 KU). The standard derivation for these values was found to be more than
double in MSG. Also Piv was measured significantly (P < 0.001) higher with MSG compared with filtrass, giving values of 21.2 ± 8.1 and 15.1 ± 7.1 mmHg, respectively. Because we could not readily explain these differences but already have shown the importance of the calibration, we grouped all MSG and their depending filtrass measurements according to their MSG calibration factor as defined above.

Differences in Kf and Piv after grouping of the data depending on the calibration factor. The pre-tensioning procedures used gave rise to good calibration factors in 21 measurements made in 16 volunteers and bad calibration factors in 29 measurements made in 20 subjects. The Kf values found with MSG with a good calibration signal were 3.3 ± 0.8 KfU, which was not significantly different from the filtrass value of 3.0 ± 0.7 KfU (Fig. 4A). In contrast, the MSG values obtained from a bad calibration signal were 5.6 ± 2.0 KfU, which was significantly (P < 0.001) greater than the value 3.3 ± 0.9 KfU measured with filtrass (Fig. 4B). There was no significant difference in the Kf values found with filtrass when the data were grouped according to the quality of the MSG calibration, whereas the MSG-based Kf was significantly higher in values with a calibration factor < 0.55 or > 1.11 (P < 0.001).

Fig. 3. A: original record from 1 typical experiment in which 11 calibration procedures on a human limb were performed for 1 mercury-in-Silastic strain-gauge plethysmography (MSG) holder. A SD of ±137 μm for the deviation of the calibration runs from the first calibration was measured in this case. B: results of a typical experiment in which 11 calibration procedures on a human limb were performed for 1 filtrass sensor. The SD of the deviations from the reference calibration was ±68 μm.

Fig. 4. A: individual (○) and mean ± SD (●) values of fluid filtration capacity (Kf) for all subjects with a good MSG calibration signal (21 observations in 16 subjects) compared with those obtained with filtrass. There was no significant difference in the mean values. B: individual (○) and mean ± SD values (●) of Kf for all subjects with a bad MSG calibration signal (21 observations in 16 subjects) compared with those obtained with filtrass. The MSG values were significantly higher than those found with filtrass (*P < 0.001).
Similarly, Piv was not significantly different when MSG results were obtained by a good calibration signal (MSG = 18.6 ± 7.1 mmHg; filtrass 15.9 ± 6.9 mmHg) but was significantly higher (P < 0.001) with a bad calibration signal (MSG 23.1 ± 8.4 mmHg; filtrass 14.4 ± 6.9 mmHg).

DISCUSSION

This paper describes a novel, automatically calibrating, liquid metal-free measuring device for VCP. The studies showed that it gave more reproducible results than simultaneous studies using MSG sensors. In our opinion, several factors may have contributed to these observations, including the reproducibility of the manufacturing process of the sensor resulting from an industrial production, the automated calibration procedure, the absence of an elastic component, and the mercury-free design of the filtrass sensor.

The MSG holders and sensors used in the present and previous studies were custom made in our own laboratories; hence differences existed in the actual design and the mechanics for calibration. This may partially account for the different mean and median $K_f$ values reported by the groups that have used this type of MSG (Table 1). The filtrass sensor, however, is an industrial product, with less variation in the sensor's mechanics. The manual calibration procedure of MSG has to be less accurate than that conducted by a motor with a position accuracy of ±5 μm. This expectation was confirmed by the calibration data obtained from the studies on both model and human limbs. The fact that filtrass utilizes the actual calibration curve as an "ideal standard," rather than a constant factor for the calculation of changes in limb volume as used in MSG, further adds to the higher accuracy of filtrass.

In MSG, silicon tubing is used in the construction of the gauges. Because the silicone material is an elastomer, it has recoil properties that, although they form an important part of the gauge function, are subject to changes attributable to altered manufacturing procedures, aging, and shelf life.

Other commercially available MSG systems do not give details on their manufacturing procedure for strain gauges. Because they are industrially produced, one assumes that the problems described above are less eminent. To our knowledge, no manufacturer, however, performs a further quality assurance besides a test for an electrical resistance change within the gauge, after delivery to the customer.

Quality of the MSG calibration. In the present study, we used filtrass as a control to test the working hypothesis that the large range of $K_f$ values obtained in control studies on adults, in our own and other published studies, result from inappropriate pre-tensioning of the Silastic strain gauges and the nonlinearity of the resistance change in the gauge when stretched. The significance of pre-tensioning can best be explained in terms of Hooke's law, which states that, until the elastic limit of a material is reached, strain is proportional to the applied stress. Clearly, with the elastomer comprising the MSG, this relationship cannot be presumed until there is sufficient pre-tension to remove the inherent "kinks" in the tubing. Although an undertensioned gauge will give uniform steps, they will be inappropriately small. One of the main disadvantages of the MSG system is that the appropriate pre-tension and the calibration cannot be checked online, and the accuracy of the calibration and, therefore, the success of the study depend on the experience of the operator. In conclusion, we believe that one of the major sources of error when MSG is used for the determination of change in length is the failure to identify an optimal pre-tension.

In clear contrast to MSG, the filtrass sensor seems not to be hampered by a variability in the calibration. Inaccuracies due to mercury-containing strain gauges are completely avoided with filtrass because the relationship of pull and response by the inductive transducer is linear and the motor always resets the measurement range to the optimum value. Moreover, because it utilizes the actual response of the inductive transducer to a defined pull by the motor, a deviation from this linearity, due for example to increased friction, can be detected. This would result in an increased time delay or a deviation from the ideally linear relationship between the defined pull and the response of the inductive transducer. The time delay is tested for each calibration, and only if a maximum of 500 ms is not exceeded can the study be continued. Furthermore, because three calibration runs are always performed and the deviation between two calibrations has to be <200 μm, we assumed that more reproducible results should result from the use of the filtrass system. This contention is supported by the results obtained in the present study.

Differences in $K_f$ and Piv. The significant differences of $K_f$ and Piv values and especially the wider spread of $K_f$ obtained with the MSG plethysmograph lead to more inaccurate findings and therefore will provide less likely detectable differences when VCP is used for microcirculatory monitoring. The calibration signal of the MSG plethysmograph, both in the one used as well as in all other commercially available strain-gauge systems, is not analyzed in routine use. This leads to an inaccuracy of VCP measurements, which can only be in part overcome with great experience of the investigator. However, it is reassuring that, given an experienced investigator and a good calibration, no significant differences in the mean values for $K_f$ and Piv were found in the present study. Moreover, the $K_f$ values reported here are in good agreement with recent data published by our group and by colleagues using a similar MSG design and pressure protocol (3, 4, 7–9, 11, 13–15, 21, 22). They are also within the range reported by other investigators (16, 17, 24–27), who gave values ranging from 2.6 to 4.8 $K_f$. The results from the present studies support the notion that the differences in control $K_f$ values that we have noticed in MSG studies (see Table 1) are most likely due to an insufficient pre-tensioning. This will result in a low calibration ratio and give rise to significantly higher $K_f$.
values. The use of a consistent and appropriate protocol has been shown to give rise to good longitudinal reproducibility of $K_f$ values with a coefficient of variation of 10.6% (9). Because all investigators have used a consistent pre-tensioning protocol, it may be less surprising that the standard deviation in the results of each of these studies is small, on the order of 5–10% (see Table 1).

All of these data contrast markedly with those of Lundvall and co-workers (18, 20) who, using volume plethysmography, reported $K_f$ values 10 times higher than previously reported values. Moreover, these workers applied venous compression pressures of 1.5–3.4 mmHg and argued that pressure steps $>5$ mmHg could activate the arteriovenous regulation mechanisms first described by Henriksen and Sejrsen (12). We found no evidence for such an activation, provided that the venous compression pressure steps were $\leq$10 mmHg (11). Indeed, we have found that venous congestion pressures, when applied in small cumulative steps, can be increased to within 20 mmHg of mean arterial blood pressure without causing a decrease in limb blood flow (10). It is interesting to note that Lundvall and Länne (20) determined $K_f$ after the application of hydrostatic pressures that were smaller than the expected capillary pressure, i.e., $\approx20$ mmHg (1, 23), which does raise questions about the validity of these results.

**Temperature.** Both skin and ambient temperature are possible sources of error when, in mercury-containing systems, changes in limb circumference are deduced from the change in resistance (28). An increase in temperature decreases the resistance of the mercury, thus having a major influence on the results obtained. In the MSG design used by our group, a temperature compensation is included in which a copper wire is in contact with both skin and ambient room air. Temperature changes occurring during a study, however, cannot be considered, because the temperature is adjusted for only at the beginning of a study. In our previous control studies, we always provided for constant skin temperature by means of a temperature control unit (11). This, however, is not always possible, particularly in clinical settings, e.g., on an intensive care ward (6). The filtrass sensor is independent of ambient temperature and free of this possible setback. Nevertheless, changes in ambient temperature should be avoided, because they have a profound influence on the measured microcirculatory parameters (19, 25).

**Comparison to other commercially available strain-gauge systems.** In this study no comparisons to other commercially available strain-gauge systems were made; hence no conclusions can be drawn with respect to the data given. However, one might speculate that the problems of MSG described in the present paper are of relevance to all MSG systems, in particular because either no calibration is performed or the quality of the calibration is not tested for. MSG plethysmographs are widely used for the assessment of limb blood flow and venous dynamics. In a yet-unpublished study, we found a significantly lower standard deviation of blood flow values with filtrass (10.3% during three repeated measurements) than with a commercially available strain-gauge system (22.3%), thus confirming our assumption that filtrass may be superior to MSG because of the improved calibration procedure and the avoidance of mercury.

**Possible drawbacks of filtrass.** The specific design of the filtrass sensor may add one source of error not present in MSG. Before any movement of the inductive transducer is initiated, the sticking friction of the plastic wire and of the friction within the inductive transducer has to be overcome. During any movement the gliding friction is always present. In contrast, the MSG behaves like a spring with very little sticking friction but, however, with a similar gliding friction. The volume responses obtained with MSG closely follow the stress-strain function of the elastic component of the silicon tubing, whereas in filtrass a step function may be seen, in particular when the sticking friction is relatively high. Although this is a potential drawback, in reality it is of minor practical importance because, during the calibration, the system automatically tests for the presence of friction, which is reflected as an increased time delay during the calibration procedure.

In conclusion, we have developed a new, automated venous compression plethysmography system and compared its quality of calibration and resulting data to those obtained using our earlier conventional MSG plethysmography. We found that the new device had smaller standard deviation during repetitive calibrations on both model and human limbs. Using the previously described protocol to determine $K_f$ and $P_{iv}$, we found that the new device gave similar values for $K_f$ when the calibration procedure of MSG was performed within 1.5 SDs of the optimum; however, incorrectly higher values were obtained with MSG when outside of this range. Because the standard deviations during repetitive measurements using the filtrass device were smaller and also because the new device utilizes the deviations of calibration curve rather than an “ideal standard” value for the calculation of changes in limb volume, we conclude that it has a higher accuracy than conventional MSG systems.

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F. Christ as an advisor and co-founder of DOMED Medizintechnik would like to disclose his commercial interest in filtrass.

This publication contains part of the MD thesis of A. Bauer.

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