Validation of the Conductance Catheter Method to Evaluate Dynamics of a Biomechanical Heart

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Abstract

The conductance catheter method has been applied to evaluate dynamics of a heart cycle. Its validation for the functional analysis of Biomechanical Hearts (BMH), autologous muscular blood pumps with a stabilizing pumping chamber, has not yet been reported. In a test apparatus, a fluid filled pumping chamber of a BMH was hydraulically compressed by a moving piston. Volume shift was measured simultaneously by a conductance catheter and by a movement analyzer as control with variation of catheter's electrical field, the pumping chamber's surrounding conductivity and the piston velocity. For all these conditions, no significant influence on the measurement accuracy was observed ($p>0.4, p>0.33, p>0.18$). Volumes measured by both methods showed a maximum relative error of 6.1±0.4% ($R=0.99$). Peak flows however demonstrated a remarkably high relative error up to 48% ($R=0.955$). Aberrations in peak flows as well as in volumes may be compensated by computed calibration formulae. The conductance catheter method offers a valuable tool for the evaluation of Biomechanical Heart dynamic properties. *(160 W).*

Key words: Biomechanical Heart, Cardiac assist device, Conductance method, Skeletal muscle ventricle, volume evaluation.

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Biomechanical Hearts (BMH) are autologous muscular blood pumps performed in a one-step operation and trained within circulation under support of drugs. A BMH consists of a skeletal muscle (musculus latissimus dorsi, MLD) and a stabilizing pumping chamber with two vascular prostheses for circulatory integration [18, 19]. The pumping chamber is made of a double layered polyurethane membrane and contains a total volume of about 145ml. These experimentally performed devices were applied in adult boor goats as chronic animal study. During operative procedure, the muscle was dissected free and wrapped around the barrel-shaped pumping chamber, which was then connected to the descending aorta (Fig. 1A). Muscle contractions were induced by a myostimulator delivering an ECG-triggered electrical stimulus via muscular implanted stimulation electrodes (4-10 pulses of 150µs pulse width, pulse frequency of 33Hz, stimulation amplitude of 2-5V).

In order to evaluate the pumping performance of a BMH such as the stroke volume and the peak ejection respectively filling rate, a reliable and practicable method of data acquisition was needed.

Conductance catheter method

Originally, the conductance catheter method was developed to evaluate the volume changes of the heart's left ventricle, described in detail elsewhere [1-4]. In brief, the technique is based on the measurement of the electrical conductivity $G(t)$ of the blood volume within the left ventricle. Therefore, a multiple electrode catheter (conductance catheter) is placed at the long axis of the left ventricle during heart catheterisation delivering an alternating current between the most proximal and distal electrodes. By means of 6 additional catheter-mounted electrodes, the electrical conductivity $G(t)$ of the surrounding blood volume was measured.

The accuracy and reliability of the conductance catheter applied within the left heart ventricle has been documented in animal experiments and in clinical practice under various conditions concerning different kinds of electrical fields, contraction velocities, ventricular sizes and stroke volumes [1-4]. Furthermore, in contrast to only experimentally implantable flow probes or microcrystals, no permanent skin perforation is necessary. This might reduce the risk of infection in chronic animal studies. Additionally, the conductance catheter method is less costly than a telemetric system. Thus, the conductance catheter method seems to be practicable to apply within the pumping chamber of a Biomechanical Heart.

To apply the conductance catheter method in Biomechanical Heart's, the multiple electrode catheter is to place inside the pumping chamber (Fig. 1B). Compared with the left heart ventricle, the pumping chamber with its electrically isolating membrane of polyurethane showed a different geometry and size. During muscle contraction it might develop irregularly, non concentric deformation. The geometric differences, the electrical isolation of the polyurethane membrane and the wide range of contraction velocities of the muscle required a validation of the conductance catheter method within BMH's ventricular cavum.

Material and Methods

In order to determine the measurement accuracy of the conductance catheter method, a mock system was to built. It should be able to simulate muscle contractions within a defined range of stroke volume with adjustable contraction velocity.

The test apparatus (Fig. 2) consisted of a fluid filled glass sphere (a), holding the saline (NaCl 0.9%) filled pumping chamber of the Biomechanical Heart (b). On its right side a compliant bladder (c) was fixed to mimic the compliance of the arterial vascular system. On the opposite side the conductance catheter entered the pumping chamber (d). The cavum of the glass sphere (a) was connected with a non compliant fluid filled tube (D_{tube} = 34 mm) leading to a movable piston (e). The piston was driven by a pneumatic pump (Festo, DSNU-16-100 PPV, Esslingen , Germany) shifting fluid from the tube into the glass sphere, which let to a compression of the pumping chamber. The movement of the piston was measured by an incremental movement analyzer (Megatron, GSM 100, München, Germany, resolution Δs_{min} = 0.056 mm, $\Rightarrow$ ΔV_{min} =

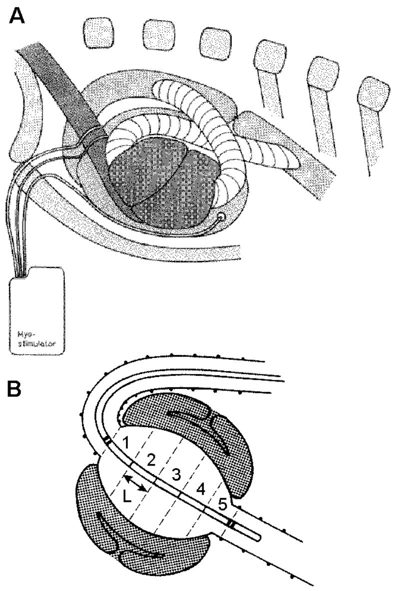

Figure 1. A: Biomechanical Heart: an autologous muscular blood pump with a stabilizing pumping chamber. The vascular prostheses are anastomosed to the descending aorta. The aorta is completely ligated between the two anastomoses. It is created during a one step operation and the muscle is trained in circulation. B: The conductance catheter is placed at the long axis of the stabilizing inlay of a Biomechanical Heart. The outer electrodes provide an alternating current. The 6 segmental electrodes allow an evaluation of conductivity between 5 segments

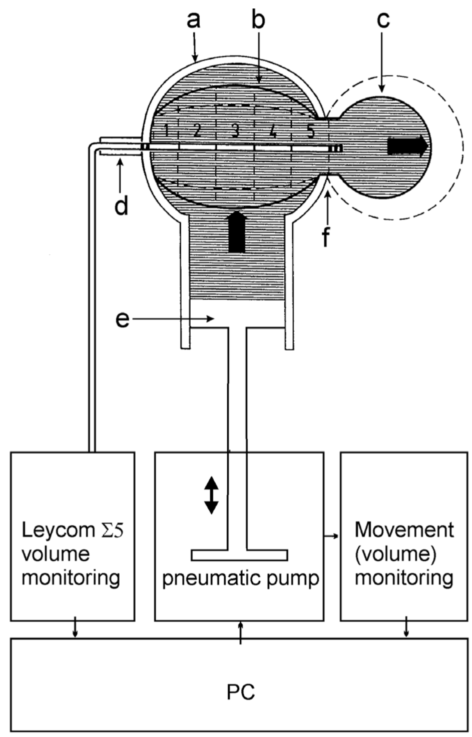

Figure 2. A mock system to simulate contractions of a Biomechanical Heart used to evaluate the measurement accuracy of the conductance method. It consists of a fluid filled glass sphere (a) holding the saline filled pumping chamber of a BMH (b). The chamber is extended with a compliant bladder (c) on the right side. A conductance catheter enters the inlay on the left side (d). A moveable piston (e) shifts volume inside the non compliant glass sphere ($V_{glass\ sphere}$ = constant), inducing a volume decrease of the BMH's pumping chamber. An ultrasonic flow probe located at the outflow tract of the pumping chamber(f) measures the outflow. Volume, flow and piston movement data are continuously digitized and stored in a personal computer. The PC also triggers the pneumatic pump.

0.051 ml). In order to examine the similarity of the volume entering the glass sphere and leaving the BMH's pumping chamber into the side bladder, a comparison of the inflow with the outflow measured by means of an ultrasonic flow probe was performed (Transonic HT207, H14, low pass filter: 30 Hz). The flow probe was located between the pumping chamber and the side bladder (Fig. 2(f)). Stroke volume adjustment was performed defining the piston's start and end position by studs. The velocity of the piston was modified by the driver pressure of the pneumatic pump altered between 4 and 6 bar. Conductance measurements were performed by a Leycom Sigma-5 signal conditioner (CardioDynamics, Leiden, The Netherlands) and a 7F conductance catheter (Cordis Europe, Roden, The Netherlands) with a distance of L = 12 mm between the segmental electrodes. The position of the catheter was central inside the lumen of the pumping chamber with the electrodes placed in the center of the long axis. The conductance catheter was used with an electrode configuration of a „large setting", this mean that the distance between the segmental electrode no. 1, 2, 5 was 12 mm and between the segmental electrode 3 and 4 24 mm. The sensitive length of the conductance catheter was thereby 84 mm. During measurement, movement analyzer's and conductance catheter's data as well as flowmeter 's data were digitized with a sample frequency of 200 Hz (AD-Converter Keithley, AD 1602, 12 Bit, Taunton, MA, USA) and stored in a personal computer (AT 486, 33 MHz).

The pumping chamber of the BMH with its barrel-shaped geometry had a maximum diameter of 60 mm, a minimum of 16 mm and a length of 90 mm. The total volume of the pumping chamber was calculated as 145 ml. Due to a sensitive length of the conductance catheter of only 84 mm but a pumping chamber length of 90 mm, the volume V_0 which surrounded the catheter between the field electrodes was to determine. This was performed by a 10 times repeated filling procedure with saline solution. The volume between the proximal and distal field electrode was determined to 139.3 ± 1.3 ml (n=10). The volume outside the sensitive part of the catheter (5.4 ± 1.4 ml) was regarded as constant and neglected over the period of a contraction.

Test procedure

Conductance catheter 's tests had to be performed 1.) regarding the variation of catheter's electrical field, 2.) investigating the presents of a parallel conductance and 3.) evaluation the influence of different contraction velocities on the measurement accuracy. In all settings, the stroke volume validation had to cover the range of 2 to 60 ml. On each setting 10 contractions were performed to evaluate the reproducibility of the conductance catheter method.

Catheter's electrical field

The electrical field was established either in a single field mode (SF) or in a dual field mode with a ratio of the outer to inner current of 0.25 and 0.3 (DF 0.25 respectively DF 0.3) and a frequency of 20 kHz. In the dual field mode, a second electrical field covered the first electrical field. This was supposed to optimize the measurement accuracy in large ventricles [4].

Parallel Conductance

The space between the glass sphere and the pumping chamber was filled with two different fluids. In the first setting, a non conductive fluid (distillate water) was used. In the second setting, the space within the glass sphere was filled with a high conductive medium (hypertonic saline solution 10%). If a parallel conductance occurred, it should be appear in the second setting as an extraventriclur conductivity leading to an overestimation of volume.

Contraction velocity

The piston's velocity was varied using a driver pressure of the pneumatic pump of 4 and 6 bar.

Analysis and statistical methods

Volumes determined with the conductance method were normalized to 139.3 ml assuming that the parallel conductance $G^P = 0$. Values were shown as mean ± standard deviation (SD). The absolute error was calculated as the difference of the conductance catheter's minus the movement analyzer's data. The relative error was computed by relating the absolute error to the corresponding data of the movement analyzer.

The effect of the electrical field (SF vs. DF 0.25 vs. DF 0.3), the filling medium of the glass sphere (non conductive vs. conductive fluid) and the contraction velocity (driver pressure: 4bar vs. 6bar) on the absolute error were statistically examined with the Wilcoxon-test (Software: WinStat 3.0, Kalmia Co, Cambridge, MA, USA). Calculated p-values smaller than 0.05 were considered as significant.

Volume data measured with the conductance catheter and the movement analyzer were correlated and a regression was performed. Peak ejection rate (PER) and peak filling rate (PFR) were derived (dV/dt) from the volume data by numerical differentiation. The peak rates of the conductance catheter's and movement analyzer's data were also correlated and a regression was computed.

Results

Fig. 3 shows the traces of the inflow into the glass sphere due to piston movement (dotted line) and outflow from the pumping chamber into the side bladder measured by means of an ultrasonic flow probe (solid line) during simulated contractions. In the period of ejection and filling, the flow values of both

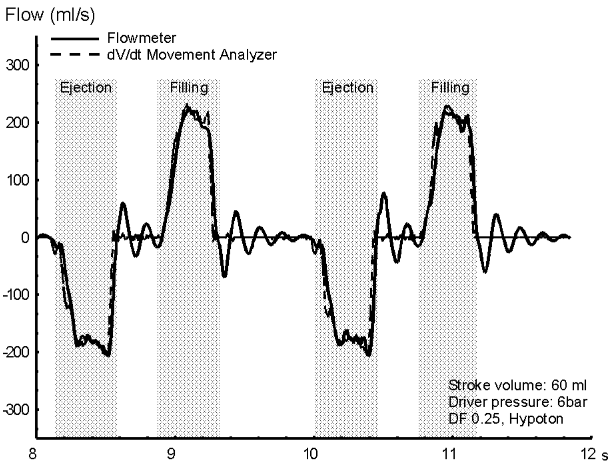

Figure 3. Tracings of movement analyzer derived flow and values measured by the flow meter. During the ejection and filling phase (grey area), no significant differences were found.

determination methods correlates with R= 0.976 (n=158). The maximum relative error was calculated to 7 %.

In fig. 4 characteristic traces of volume changes (left) related to the pumping volume V_0 =139 ml and flow (right) are shown. The shifted volume was 5ml (top), 30 ml (middle) and 60 ml (bottom). During volume evaluation with both methods, stroke volumes of 5ml and 60 ml resulted in an absolute error of 1.2ml respectively -0.8 ml, while a stroke volume of 30 ml differed in - 6.1 ml. This finding let expect a non linear regression between both methods. Evaluating peak ejection and filling rates, it was to consider that in addition to an increased driver pressure, the velocity of the piston was not linear during an increasing stroke volume. In the stroke volume range of 5 – 30 ml, the peak rate reached values of 100 ml/s up to 350ml/s. Beyond that stroke volume no additional increase of the piston velocity happened; peak ejection and filling rates were nearly constant.

Regarding 80 simulated contractions within a stroke volume range of 2 to 60 ml, an absolute error of 1.3 ± 0.4 ml (stroke volume between 2-10 ml and 50 – 60 ml) and – 5.4 ± 1.2 (stroke volume between 20 – 40 ml) was calculated (electrical field: DF0.25, glass sphere filling: distilled water and driver pressure: 6bar). The SD of 10 subsequent measured contractions was mostly in the range of 0.2 – 0.3 ml. Only in the stroke volume range of 20 – 40 ml, the SD rose up to 1.2 ml. The SD of the movement analyzer's data was always in the range of 0.1 – 0.2 ml.

Electrical field in single and dual field mode

Comparing the absolute error of the volume data at variation of the electrical field (n=108) in the modes SF, DF 0.25 and DF 0.3, no significant differences were found (0.18 < p < 0,42) (Tab. I). This demonstrates that

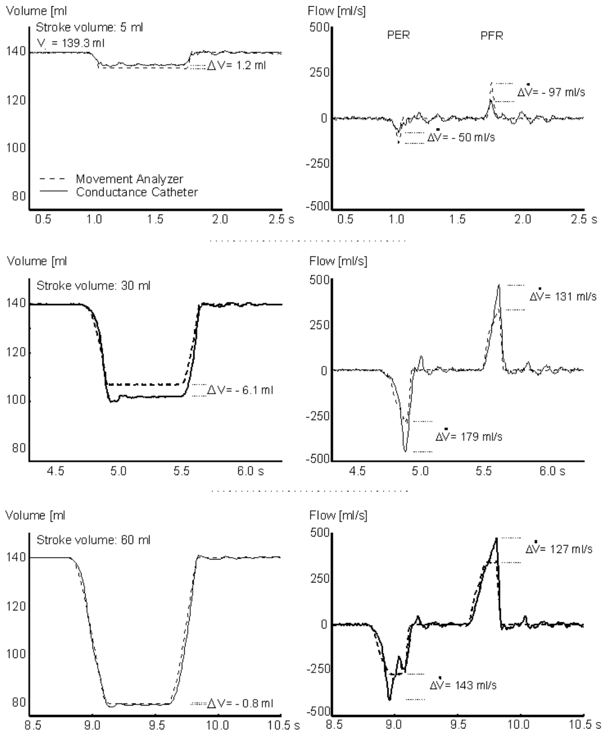

Figure. 4: Tracings of simulated contractions with stroke volumes of 5, 30 and 60 ml. On the left side, the volume decrease of the pumping chamber (solid line) due to the piston's volume displacement (dotted line) is shown. On the right side, the derivations of volume data with peak ejection and filling rates PER resp. PFR are presented.

the measurement accuracy of volume does not depend on the mode of electrical field in our test setup.

Parallel Conductance

Regarding the two different fillings of the glass sphere either with a non conductive or a high conductive fluid, the determined volumes of the conductance catheter and movement analyzer showed no significant differences (n=142, p=0,19, Tab. I). Thus, the parallel conductance had no influence on the measurement accuracy of the conductance catheter method in our test setup. Consequently it can be neglected.

Contraction velocity

Piston's velocity depended on one hand on the driver pressure (4bar, 6bar) and on the other hand on the acceleration duration of the piston. At small stroke volumes, the piston could not reach the maximum velocity due to short acceleration time, but at larger stroke volumes, the maximum piston velocity was gained as a force balance between driver pressure versus piston friction and increasing system pressure (max. pressure 100mmHg). This resulted in a wide range of piston velocity and flows between 50 ml/s (stroke volume: 2ml, driver pressure: 4 bar) and 350 ml/s (stroke volume > 30 – 60 ml, driver pressure: 6 bar).

Table 1: Comparison of the absolute error under different test conditions

	Test condition	Constant
n= 108	Kind of electrical field *SF ⇔ DF 0.25 ⇔ DF 0.3* └ p= 0.18 ┘ └ p= 0.34 ┘ └──── p= 0.42 ────┘	non conductive fluid 4 bar
n= 142	Parallel Conductance *Non conductive ⇔ conductive fluid* └──── p= 0.19 ────┘	DF 0.25 4 bar
n= 93	Velocity (Driver Pressure) *4 bar ⇔ 6 bar* └ p= 0.13 ┘	non conductive fluid DF 0.25

Evaluating the absolute error of the volume determined under the aspect of the two different driver pressures (4 bar, 6 bar), no significant differences were found (n=93, p=0.13). This means that independently to the simulated contraction velocity, the absolute error of volume was at the same level.

These statistical evaluations demonstrate that neither the type of electrical field (SF ⇔ DF 0.25 ⇔ DF 0.3) nor the conductivity of the surrounding of the pumping chamber (distillate water ⇔ saline solution NaCl 10%) and the piston's driver pressure (4 bar ⇔ 6 bar) had any influence on the absolute error of the volume evaluation. Taking this into account, the whole volume data set could be merged to calculate the correlation between movement analyzer and conductance catheter's data.

Volume

Figure 5 demonstrates the merged volume data fitted with a polynomial regression of 2nd order between the conductance catheter and movement analyzer's determined volume. The correlation coefficient between both volume determination methods was calculated to R = 0.9986 (n=343). Regarding the point of interception between the regression curve and the line of identity, the conductance catheter method overestimated the volume changes in the measured range of 77 to 132 ml. Thereby, the end systolic volume during simulated contraction was evaluated smaller than it was. The largest differences between both methods was calculated at pumping chamber volumes of 110 ml (stroke volume about 30 ml). The absolute error was calculated between –4.6 to –7.3 ml. This corresponds to a maximum relative error of 6.6% related to the pumping chamber volume of 110 ml. If instead the absolute error was related to the stroke volume (SV: 30 ml), the maximum relative error was calculated to 24.3%.

In order to correct the data of the conductance catheter method, the overestimation had to be subtracted: $V_{\text{corr.}} = V_{\text{Conductance}} - \Delta \text{V}$. This let to a calibration formula V described by:

$$V = V_C - \Delta V$$

$$\Delta V = 4.73 \cdot 10^{-3} \cdot V_C^2 - 0.99 \cdot V_C + 47.7 \text{ ml}$$

$$V = -4.73 \cdot 10^{-3} \cdot V_C^2 + 1.99 \cdot V_C - 47.7 \text{ ml}$$

Fig. 7, top demonstrates the effect of the calibration on the absolute error of the conductance catheter's volume evaluation. Without calibration, the absolute error is shown by the light circles and fitted to the polynomial curve. After calibration, the absolute error (shown with

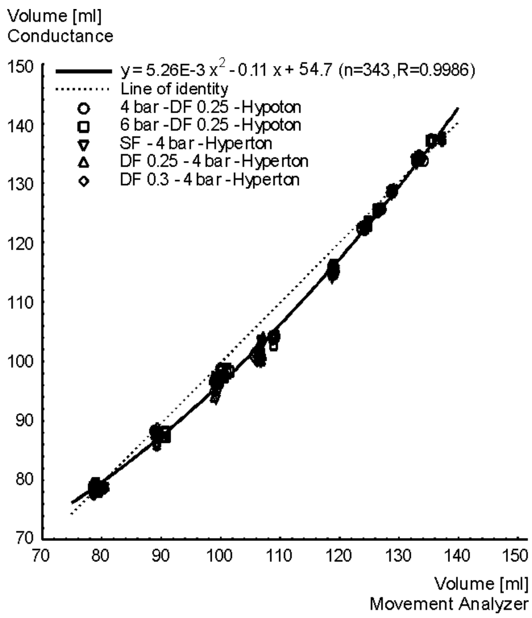

Figure 5. Simulated contractions with different test conditions correlates with a coefficient of R=0.9986. The regression fits a polynom of 2nd order. Values below the line of identity (dotted line) indicates an overestimation of stroke volume; the end systolic volume was evaluated smaller than it was

dark circles) is constantly smaller than ± 2ml (Fig. 7, top, grey area).

Peak Ejection Rate, Peak Filling Rate

Peak ejection rate (PER) and peak filling rate (PFR), the derivate quantities of the measured volume, showed a correlation of R = 0.955 between conductance catheter's and movement analyzer's data in the validation range of 75 to 400 ml/s (n=242). Regarding the absolute error of the peak ejection versus peak filling rate, no significant differences (p = 0.11) could be found. Thereby the ejection and filling peak rates could be combined to a merged peak rate. Fig. 6 demonstrates the linear regression of the peak rates. Calculation of the absolute error resulted in an underestimation of the peak rate in the measured range of 50 to 120 ml/s and an overestimation between 120 ml/s and 550ml/s. In fact, the maximum peak rate measured by means of the movement analyzer was 370 ml/s. This corresponds to a maximum absolute error of 180 ml/s. The maximum relative error was 48.6 %. In order to diminish these aberrations, a calibration formula was calculated by means of the reverse function with:

$$PR = PR_C - \Delta PR$$
$$\Delta PR = 0.37 \cdot PR_C - 35.4 \text{ ml/s}$$
$$PR = 0.63 \cdot PR_C + 35.4 \text{ ml/s}$$

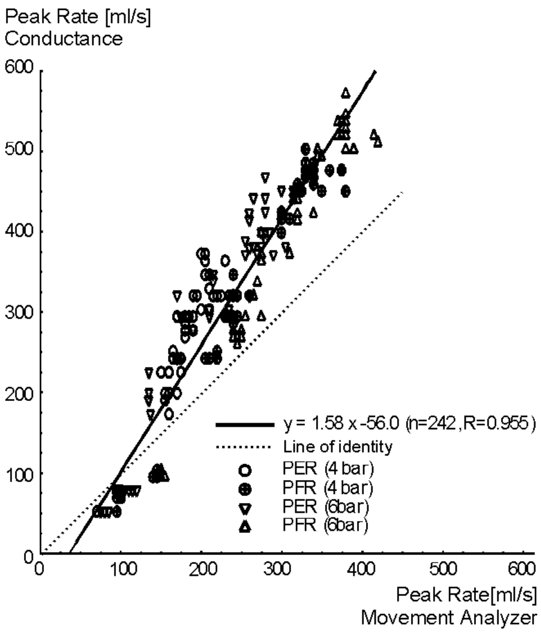

Figure 6. Linear regression of the volume derivated peak ejection and filling rate (PER/PFR). The y-axis represents the conductance measured peak rates, the x-axis shows the given peak rates performed by the piston's movement. The circle data show the peak rates at 4 bar and stroke volume at 2 ml and 60 ml. The triangle data represents the value at 6 bar driver pressure of the pneumatic pump.

Fig. 7, bottom shows the difference of the original (light circle) and calibrated (dark circle) measurements. The absolute error of the original peak rates ranged between –50 ml/s up to 150 ml/s. After calibration, the values were constantly smaller than ± 30 ml/s (grey area).

Discussion

The conductance catheter method is an experimentally and clinically established procedure to evaluate left ventricular's volume within the beating heart. Therefore, the electrical conductivity of the intraventricular blood volume, a scaling factor α and the parallel conductance Pc has to be determined. Evaluation of these parameters gave sources of mismeasurement and inaccuracy. However, if the conductance catheter method was applied within the pumping chamber of a BMH, neither the scaling factor α nor the parallel conductance Pc had to be determine. That is because, the scaling factor α could be calculated by relating the maximum of the conductance catheter measured volume to the real pumping chamber volume V_0. Additionally, the parallel conductance Pc could be neglected due to the electrical isolation of the pumping chamber to the surrounding tissue.

Regarding the remarkable relative error of about 48% concerning the evaluation of the peak rates, it is to consider that PER and PFR were gained by numerical differentiation of the volume data. These volume data fitted to a polynom of 2nd order. By differentiating a polynomial function of 2nd order, one achieves a linear

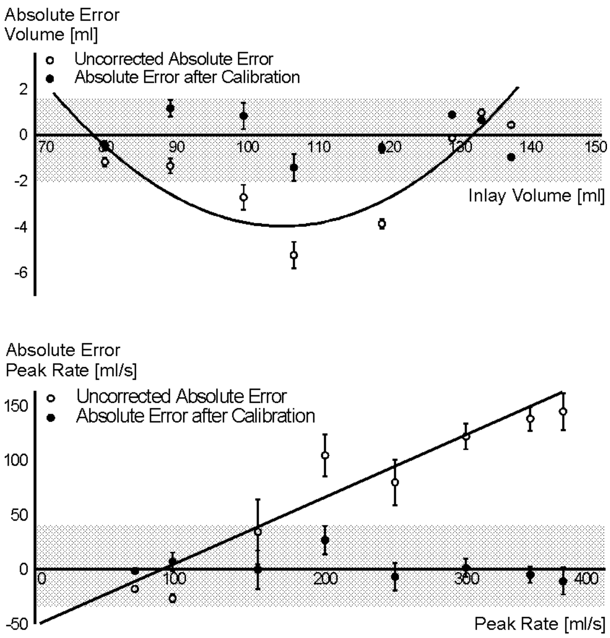

Figure 7. Absolute error of volume (top) and peak rate (bottom). Without any calibration, the absolute error fitted the drawn curves. Using the calibration formulae, a reduction of the absolute error within the grey area was achieved.

function with a two times increased factor and a loss of constants. Thus, relating the absolute error of the volume evaluation to the stroke volume instead of the initial volume, a maximum relative error of about 23 % was calculated (s.o). This was roughly the half of the observe relative error of the peak rates. But nevertheless, with the calibrated peak rates, an evaluation of the dynamic muscle properties was possible.

Comparing the conductance catheter method with other measurement techniques like a transthoracic ultrasound evaluation or an x-ray examination, it is to consider, that the conductance catheter method has a at least 10 time higher time resolution. This is important to evaluate the dynamics of our muscle pump during each part of a contraction cycle. Additionally, the measurement accuracy of these picture based techniques are limited, if the muscle contraction is not regular and concentric.

An in vivo transfer of the conductance catheter method seems feasible. However, while the similar dimensions of the pumping chamber and its electrical properties have to be employed. Even, if the measurement of the scaling factor α and the parallel conductance Pc is not necessary in our test setup, it has to regard their influence in an in vivo application concerning muscle constriction and biological coating (fibrin) to achieve high measuring accuracy. In combination with a pressure recording within the pumping chamber via a tip mounted pressure module, pressure-volume loops can be constructed. With this additional information the dynamic pressure development and the stroke work as area inside the PV-loops can be determined.

In conclusion, the conduction catheter method offers a reliable and valuable tool to evaluate dynamics of Biomechanical Hearts.

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