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## Comparison of MRI, 64-slice MDCT and DSCT in assessing functional cardiac parameters of a moving heart phantom

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**Abstract** To compare magnetic resonance imaging (MRI), 64-slice multi-detector computed tomography (MDCT) and dual-source computed tomography (DSCT) in assessing global function parameters using a moving heart phantom. A moving heart phantom with known volumes (215–258 ml) moving at 50–100 beats per minute was examined by three different imaging modalities using clinically implemented scanning protocols. End-diastolic and end-systolic volumes were calculated by two experienced observers using dedicated post-processing tools. Ejection fraction (EF) and cardiac output (CO) were calculated and mutually compared using Bland-Altman plots. MRI underestimated the ejection EF by 16.1% with a Bland-Altman interval (B-A) of  $[-4.35 (-2.48) -0.60]$ . Sixty-four-slice MDCT overestimated the EF by 2.6% with a relatively wide B-A interval of  $[-3.40 (0.40) 4.20]$ .

DSCT deviated the least from the known phantom volumes, underestimating the volumes by 0.8% with a B-A interval of  $[-1.17 (-0.13) 0.91]$ . CO analysis showed similar results. Furthermore, a good correlation was found between DSCT and MRI for EF and CO results. MRI systematically underestimates functional cardiac parameters, ejection fraction and cardiac output of a moving heart phantom. Sixty-four-slice MDCT underestimates or overestimates these functional parameters depending on the heart rate because of limited spatial resolution. DSCT deviates the least from these functional parameters compared to MRI, EBT and 64-slice MDCT.

**Keywords** Functional parameters · MRI · Sixty-four-slice MDCT · DSCT · Phantom · Global cardiac function

### Introduction

Functional parameters of the left ventricle, such as the ejection fraction and the cardiac output, are closely related to cardiac morbidity and mortality. These parameters can predict the prognosis of coronary artery disease in individual patients and in entire cohorts if assessed accurately [1–4]. The current standard of reference to assess functional parameters is short-axis magnetic resonance imaging (MRI) using steady-state free precession (SSFP) sequences [5–10]. The main advantage of MRI is its excellent temporal resolution without the need to expose

the patient to ionising radiation or nephrotoxic contrast agents. However, MRI is contra-indicated in a substantial number of patients for various reasons, e.g., non-MR-compatible implants or claustrophobia [11].

As an alternative to MRI, multi-detector computed tomography (MDCT) can be used to assess functional parameters of the left ventricle. Although MDCT has a lower temporal resolution (165 ms), its spatial resolution (voxel size  $0.6 \times 0.6 \times 3.0$  mm) is higher than MRI (voxel size  $1.7 \times 1.7 \times 6.0$  mm). Another CT technique, dual-source CT (DSCT), has a temporal resolution more similar to the temporal resolution of MRI (83 ms). DSCT combines this

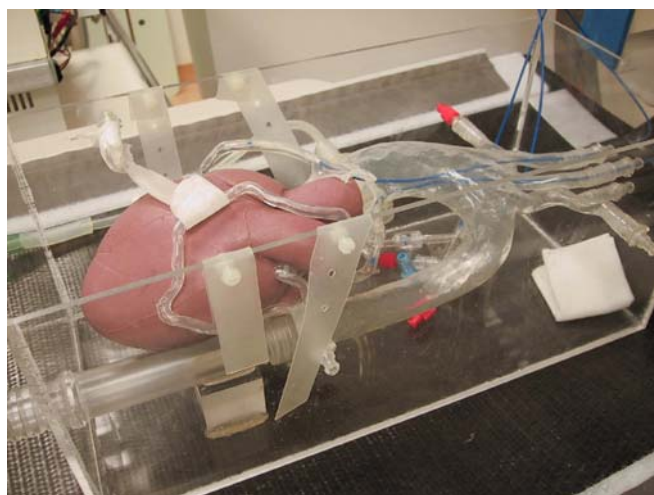
relatively high temporal resolution with a relatively high spatial resolution (voxel size  $0.6 \times 0.6 \times 3.0$  mm) [12]. These given resolutions are based upon clinically implemented scan protocols. Although these resolutions can be further optimised, this could influence negatively other aspects of the scan, such as scan duration, patient dose or noise levels.

Since CT is now used more frequently in clinical practice than MRI, it is relevant to assess its diagnostic accuracy for analysis of functional parameters of the left ventricle, since these parameters can easily be calculated from the raw data of a gated CT examination of the heart. Whereas CT is the preferred modality for assessing morphology of the heart and the coronary arteries, CT might also play an important role in functional assessment as well.

Therefore, the purpose of this study was to compare MRI, 64-slice MDCT and DSCT in assessing the ejection fraction and cardiac output of an anthropomorphic moving heart phantom at a range of different heart rates.

## Materials and methods

An anthropomorphic moving heart phantom (Limbs & Things, Bristol, UK) with known volumes was connected to a gas pump with variable flow settings (Fig. 1). The gas pump also generated a trigger signal that enabled synchronised ECG gating of the imaging modalities with the movement of the phantom. The phantom was designed and adjusted for CT and MRI applicability [13]. Heart rates of 50–100 beats per minute (bpm) with an interval of 10 bpm were simulated. The known phantom volumes were used as the reference. The reference value for the end-diastolic volume (EDV) was determined by filling the phantom with water. This value was considered to be constant at varying heart rates. The reference end-



**Fig. 1** The moving anthropomorphic heart phantom

systolic volume (ESV) was calculated by determination of the stroke volume (SV) using a gas flow analyser (VT Plus HF, Fluke, Everett, WA) and using the relation  $SV = EDV - ESV$ .

The anthropomorphic moving heart phantom was positioned along the z-axis of the scanner and scanned with three different imaging modalities using clinically implemented scanning protocols: MRI, 64-slice MDCT and DSCT. The parameters for 1.5-T MR (Siemens Magnetom Sonata, Erlangen, Germany) were: TR/TE 57.46/1.10 ms,  $\alpha$  59°, FOV 284 × 350 mm, matrix 125 × 192 mm, voxel size  $1.7 \times 1.7 \times 6$  mm and interslice gap 4 mm with a retrospectively gated cine steady-state free-precession sequence. The parameters for 64-slice MDCT (Siemens Somatom Sensation 64, Erlangen, Germany) were: rotation time 330 ms, slice thickness 3.0 mm increment 3.0 mm, tube voltage 120 kV and tube current 250 ms. The parameters for DSCT (Siemens Somatom Definition, Forchheim, Germany) were: rotation time 330 ms, slice thickness 3.0 mm, increment 3.0 mm, tube voltage 120 kV and tube current 120 mAs/rot, a protocol similar to that of the 64-slice MDCT.

After data acquisition the 64-slice MDCT and DSCT images were reconstructed with a medium smooth kernel at every 10% of the RR interval. A bi-segmental reconstruction was used for the 64-slice MDCT data for heart rates over 65 bpm as suggested by the manufacturer, and a single-segmental reconstruction was used for the DSCT data.

Then, the reconstructed axial images were analysed by two independent experienced observers. Both observers had more than 3 years of experience in analysing and post-processing clinical cardiac MRI and CT scans. Two dedicated post-processing tools, using the same algorithm and the same visual interface, were used for this analysis. The volume analysis of the MRI and DSCT data was performed using QMass (Medis, Leiden, The Netherlands). The volume analysis of the 64-slice MDCT data was performed using CardIQ (GE Medical Systems, Milwaukee, WI). In the volume analysis, the end-diastolic volume (EDV) and the end-systolic volume (ESV) were calculated. Since the phantom was scanned along the z-axis, short-axis images were obtained automatically. Apical and basal slices were defined as the first and last slice in which the ventricular lumen was present. The diastolic phase for the EDV was defined as the phase with the greatest luminal cavity. The end-systolic phase for the ESV was defined as the phase with the smallest luminal cavity. The observers manually traced the area of the cavity in each slice in the appropriate phases, defining the total volume. The obtained volumes of the analysis were used to calculate the stroke volume (SV), ejection fraction (EF) and cardiac output (CO) at every heart rate. Stroke volume is calculated as  $SV = EDV - ESV$ . The ejection fraction is calculated as  $EF = (SV/EDV) * 100\%$ . The cardiac output is calculated as  $CO = HR * SV$ .

## Data analysis

The functional parameters of the imaging modalities were compared to the functional parameters of the reference and assessed using an over- or underestimation and a relative difference. The over- or underestimation is defined as the difference in functional parameter between one of the four modalities and the reference. The relative difference is defined as the over- or underestimation in functional parameter by a modality as a percentage of the functional parameter of the reference.

The method of Bland and Altman was used to display the average over- or underestimation and limits of agreement between the imaging modalities and the reference [14]. The results are shown in the form of [*average - 2x standard deviation (average) average + 2x standard deviation*]. A small over- or underestimation indicates a high accuracy, and small limits of agreement indicate a high precision.

A statistical test was used to investigate the significance of the obtained volumes and estimations. A paired t-test was used to assess the differences in the measured volumes by the two observers. The same statistical test was used to assess the over- and underestimations in EF and CO as a function of HR by the different modalities. A p-value < 0.05 was considered to be significant for the test.

## Results

### Volumetric data

The reference maximum volume, end-diastolic volume (EDV), of the heart phantom is  $258 \pm 1$  ml. The reference minimum volume, end-systolic volume (ESV), of the heart phantom varies at different heart rates. The ESV increases with increasing heart rates, ranging from 215.0 ml at 50 bpm to 223.7 ml at 100 bpm.

Averaged EDV values were between 263.2 and 269.6 ml for MRI, between 274.6 and 283.2 ml for 64-slice MDCT and between 256.9 and 261.7 ml for DSCT, showing exaggerated EDVs compared to the reference for all

modalities except for DSCT. Averaged ESV values were between 229.0 and 236.9 ml for MRI, between 230.2 and 248.4 ml for 64-slice MDCT and between 219.8 and 220.6 ml for DSCT, showing exaggerated ESVs compared to the reference for all modalities except for DSCT again. The measured volumes are given in detail in Tables 1 and 2, and axial slices of the heart phantom obtained with the three imaging modalities are shown in Fig. 2.

The volumes were used to calculate the different functional parameters. The ejection fraction (EF) of the reference is smaller at higher heart rates, a pattern observed with the three modalities as well (Fig. 3). The EF by MRI was underestimated compared to the reference at all heart rates. The average relative difference of -16.1% to the reference for MRI was significant ( $p < 0.01$ ). The EF by 64-slice MDCT was overestimated compared to the reference except at heart rates of 60 and 100 bpm. However, the average relative difference of 2.6% to the reference was not significant ( $p = 0.62$ ). The EF by DSCT showed a small underestimation at lower heart rates and a small overestimation at higher heart rates. The average relative difference of 0.8% to the reference was not significant as well ( $p = 0.57$ ).

In contrast to the EF, the cardiac output (CO) of the reference is larger at higher heart rates, a pattern observed for all three investigations (Fig. 4). The CO by MRI was underestimated compared to the reference at all heart rates. The average relative difference of -12.7% to the reference for MRI was significant ( $p < 0.02$ ). The CO by 64-slice MDCT was overestimated compared to the reference except at heart rates of 60 and 100 bpm. However, the average relative difference of 10.7% to the reference was not significant ( $p = 0.15$ ). The CO by DSCT was underestimated except at a heart rate of 100 bpm. The average relative difference of 0.3% to the reference was not significant as well ( $p = 0.84$ ). The relative differences to the reference for each modality are given in Tables 1 and 2.

The accuracies (size of over- or underestimation) and precisions (range of limits of agreement) of the different imaging modalities in assessing the EF are shown using Bland-Altman plots (Fig. 5). The accuracy and precision of MRI were both moderate in assessing the EF. The accuracy

**Table 1** End-diastolic volumes of the moving heart phantom as measured by the different imaging modalities

HR	Ref	MRI		64-slice MDCT		DSCT	
		Obs 1	Obs 2	Obs 1	Obs 2	Obs 1	Obs 2
50	258	267.4	263.0	279.2	285.1	262.1	261.4
60	258	267.3	267.0	279.8	284.7	262.1	261.3
70	258	270.5	268.0	274.3	280.2	261.2	260.7
80	258	267.2	265.2	279.3	287.0	260.0	259.4
90	258	263.2	263.8	274.6	274.7	257.8	257.0
100	258	265.9	273.2	280.3	278.7	257.3	265.4

**Table 2** End-systolic volumes of the moving heart phantom as measured by the different imaging modalities

HR	Ref	MRI		64-slice MDCT		DSCT	
		Obs 1	Obs 2	Obs 1	Obs 2	Obs 1	Obs 2
50	215.0	233.0	229.0	220.3	219.3	228.4	232.9
60	215.9	231.4	226.7	220.3	219.3	241.9	238.2
70	216.9	235.4	228.5	221.2	219.1	235.6	230.6
80	217.6	234.2	232.4	220.8	219.2	231.2	231.7
90	220.8	230.3	230.3	221.1	219.4	232.3	228.1
100	223.7	234.0	234.0	221.8	219.4	251.0	245.9

of 64-slice MDCT was much better, but the precision was relatively low. The accuracy and precision of DSCT were both high. The precise Bland-Altman limits for the different modalities in assessing the EF were  $[-4.35 (-2.48) -0.60]$  for MRI,  $[-3.40 (0.40) 4.20]$  for MDCT and  $[-1.17 (-0.13) 0.91]$  for DSCT.

The accuracies and precisions found in the analysis of the CO results were different from the previous EF results (Fig. 6). MRI and 64-slice MDCT both showed a moderate accuracy. However, the precisions were different. The precision of MRI was high, but the precision of 64-slice MDCT was low. DSCT showed again a high accuracy and a high precision. The precise Bland-Altman limits for the different modalities in assessing the CO were  $[-0.72 (-0.37) -0.02]$  for MRI,  $[-0.58 (0.31) 1.20]$  for MDCT and  $[-0.18 (0.01) 0.20]$  for DSCT. The accuracy of each modality is given in Table 3.

#### Inter-observer data

Finally, the inter-observer analysis showed that, overall, the difference in volumes as determined by the two observers

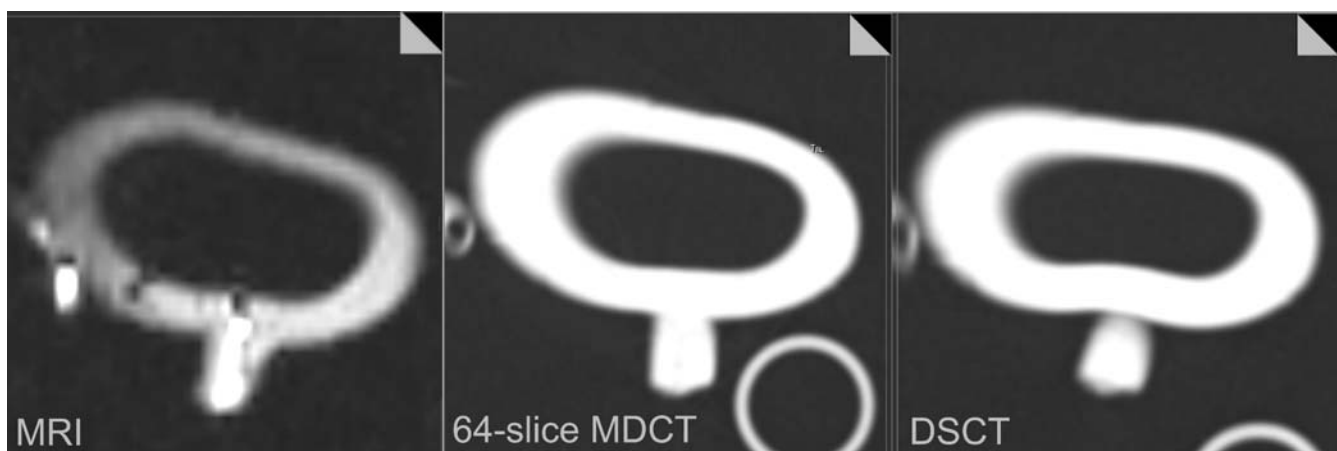
was not significantly different ( $p > 0.08$ ). However, when analysing the difference in EDV and the difference in ESV separately, the difference in EDV was not significant ( $p > 0.90$ ), whereas the difference in ESV was significant ( $p < 0.01$ ).

#### Discussion

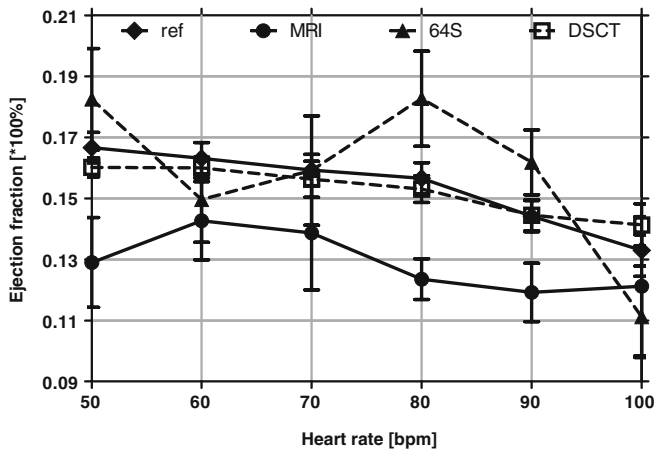
Our results show that MRI underestimates functional parameters, ejection fraction and cardiac output of a moving heart phantom. Sixty-four-slice MDCT underestimates or overestimates these parameters depending on the heart rate. DSCT deviates least from these parameters compared to MRI and 64-slice MDCT.

The different imaging modalities inevitably used slightly different scan protocols in this study. Although it would have been possible to use more similar parameters, such as comparable slice thicknesses and increments, it was considered to be more clinically relevant to use typical protocols used in daily clinical practice.

MRI overestimated the end-diastolic volume (EDV) and end-systolic volume (ESV) structurally, as Mao et al. found



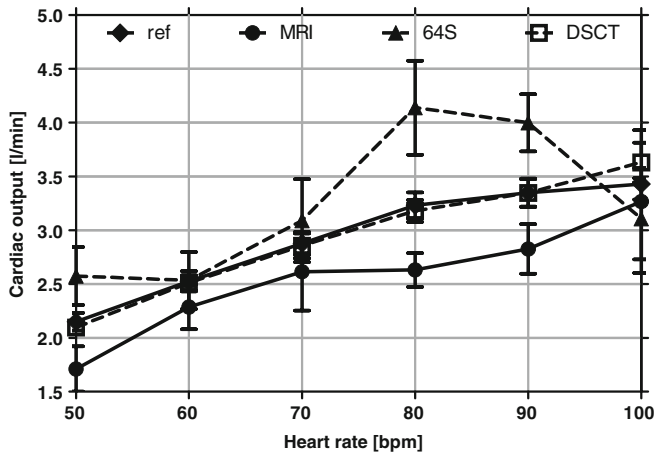
**Fig. 2** Axial slices of the heart phantom as obtained with MRI, 64-slice MDCT and DSCT



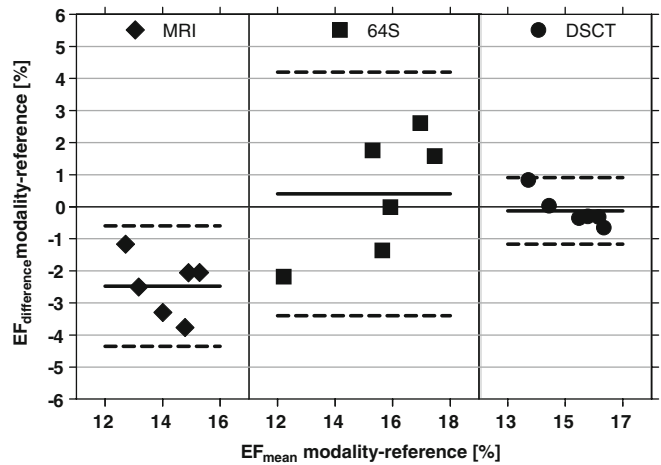
**Fig. 3** Ejection fraction (EF) as function of heart rate (HR) obtained with the reference (solid line with  $\blacklozenge$ ), MRI (solid line with  $\bullet$ ), 64-slice MDCT (dotted line with  $\blacktriangle$ ) and DSCT (dotted line with  $\square$ )

as well [15]. This resulted in a structural underestimation of the ejection fraction (EF) and cardiac output (CO). The structural overestimation of the physical volumes might be the result of the limited spatial resolution of MRI. For MRI, the section thickness was 6 mm, a reduced spatial resolution compared to the other modalities, explaining this overestimation. Sequences with higher spatial resolution are available for MRI, and it is expected that using these optimised sequences will improve the results. However, the duration of the scan will be increased.

Sixty-four-slice MDCT is not hampered by a limited spatial resolution; however, its temporal resolution is low. This low temporal resolution causes increased blurring in

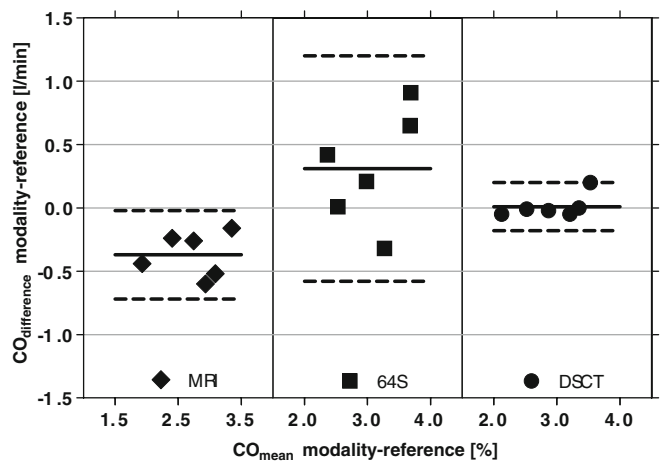


**Fig. 4** Cardiac output (CO) as function of heart rate (HR) obtained with the reference (solid line with  $\blacklozenge$ ), MRI (solid line with  $\bullet$ ), 64-slice MDCT (dotted line with  $\blacktriangle$ ) and DSCT (dotted line with  $\square$ )



**Fig. 5** Bland-Altman plots comparing the ejection fraction (EF) of the reference measurement to MRI ( $\blacklozenge$ ), 64-slice MDCT ( $\blacksquare$ ) and DSCT ( $\bullet$ ). The solid lines are the means, the dotted lines two times the standard deviation (2 SD)

the CT images, making a proper delineation of the luminal cavity more difficult. Therefore, an observer might easily over- or underestimates a volume at an individual heart rate (e.g., the rather large deviation at 80 bpm), whereas the average measurement over all heart rates has a high accuracy. This explains the variable results with large limits of agreement for 64-slice MDCT. The average measurement is accurate, but it shows large outliers. Similar results were reported previously; data acquired with a temporal resolution of 165 ms showed large limits of agreement, whereas data acquired with a temporal resolution of 83 ms showed very small limits of agreement [16]. Although the temporal resolution can be improved with



**Fig. 6** Bland-Altman plots comparing the cardiac output (CO) of the reference measurement to MRI ( $\blacklozenge$ ), 64-slice MDCT ( $\blacksquare$ ) and DSCT ( $\bullet$ ). The solid lines are the means, the dotted lines two times the standard deviation (2 SD)

**Table 3** Over- or underestimations (accuracy) and relative differences between the reference and the modalities for the ejection fraction (EF) and the cardiac output (CO) averaged over all heart rates

	MRI		64-slice MDCT		DSCT	
	Est	Rel	Est	Rel	Est	Rel
EF	-2.48	16.1	+0.4	2.6	-0.13	0.8
CO	-0.37	12.7	+0.44	15.1	+0.01	0.3

The relative differences and estimations in EF are given as a percentage. The estimation in CO is given in l/min. *Est* =over- or underestimation; *rel* =relative difference

multi-segment reconstruction, leading to better reproducibility in a phantom study [17], it was shown that this technique did not improve results for studies in human subjects [18].

DSCT with a high temporal and a high spatial resolution approximated the physical volumes the best. Another phantom study by Mahnken et al. showed also good approximations to the physical volumes with DSCT [16]. In vivo studies have also reported on the use of DSCT for functional assessment. However, these studies only result in a comparison between DSCT and usually MRI, because physical values are not known. The reported correlations between DSCT and MRI are very good [19–21].

There were no clinically significant differences in image quality for both CT systems with respect to delineation of the ventricular cavity (Fig. 2). It should be noted that the smooth surface of the ventricular cavity in the heart phantom is not necessarily representative of the situation in vivo. The human left and right ventricular cavities are lined with varying degrees of inhomogeneous porous myocardial tissue, often referred to as trabecularisation [22]. Furthermore, the contrast between blood and myocardial tissue in vivo is different than the contrast between artificial material and gas in the heart phantom. Therefore, the aspect of contrast and image quality could not be fully taken into account in the present study.

The measured EDVs and ESVs resulted in very low EFs and COs. Such low EFs and COs are uncommon in clinical practice. Besides these uncommon EFs and COs, there are two more aspects that make this phantom study deviate from clinical practice. First, the moving heart phantom has a fixed ejection fraction and cardiac output at a certain heart rate, whereas in patients the functional parameters may vary. Second is the phenomenon of trabecularisation of the left ventricle, which makes exact delineation of the endocardial border in patients more dependent on visible contrast between the blood pool and the ventricular wall as stated previously. Despite these shortcomings, the use of a moving heart phantom offered the advantage of ensuring highly reproducible conditions at each of the three imaging modalities, whereas studies in human subjects intrinsically

suffer from lack of consistency due to different variables, such as physiological non-constant heart rates caused by breathing or fluctuations in fluid status.

Although the moving heart phantom has some shortcomings and deviates from clinical practice, the results might be extrapolated to EFs and COs more common in clinical practice. The small over- and underestimations of the functional parameters of the left ventricle might argue for interchangeable results for different imaging modalities. However, these over- and underestimations are due to the fact that the simulated functional parameters were relatively small as well. If these results are extrapolated to values associated with normal EFs and COs [23, 24], the over- and underestimations might be around 10% for the EF and around 0.9 l/min for the CO.

In addition to the use of different imaging modalities resulting in different values of functional parameters, human error might increase this difference. Surprisingly, there was a significant difference in the ESV between the two observers, whereas the EDV showed no difference. These differences cannot be explained by visual selection of different phases or systematic differences in delineation. These differences are most likely explained by larger differences in terms of percentage between the intrinsically lower quantitative values for ESV compared to EDV.

### Limitations

Reconstructions were made at every 10% of the RR interval, although smaller intervals are possible. Smaller intervals, for example intervals of 5%, have been used in previous studies [25, 26]. However, reconstructions at intervals of 10% are thought to be adequate for clinical use according to Suzuki et al. [27]. In addition, it was reported that DSCT and MRI show similar differences in functional parameters when 10 or 20 phases are used for reconstruction of the DSCT data [28].

Different post-processing tools were used for the volume analysis, because the individual post-processing tools could only process CT or MR data and not both. Furthermore, not all post-processing tools were available throughout the entire study, and since the DSCT data were added in a later stadium, it could not be analysed using CardIQ. However, all post-processing tools were developed by the same company (Medis, Leiden, The Netherlands) and use the same algorithm and visual interface. Therefore, the differences in end-diastolic and end-systolic volumes are not considered to be influenced by using the different post-processing tools.

### Conclusion

Our results show that a clinically implemented MRI protocol structurally underestimates functional parameters,

ejection fraction and cardiac output of a moving heart phantom. A clinical implemented protocol using 64-slice MDCT underestimates or overestimates these functional parameters depending on the heart rate. A clinical protocol using DSCT deviates the least from these functional parameters compared to MRI and 64-slice MDCT.

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