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FULL PAPER

Influence of cardiac motion on stent lumen visualization in third generation dual-source CT employing a pulsatile heart model

¹NILS PETRI, MD, ²TOBIAS GASSENMAIER, MD, ³THOMAS ALLMENDINGER, PhD, ³THOMAS FLOHR, PhD, ¹WOLFRAM VOELKER, MD and ²THORSTEN A BLEY, MD

¹Department of Internal Medicine I, University Hospital of Würzburg, Würzburg, Germany

²Department of Diagnostic and Interventional Radiology, University Hospital of Würzburg, Würzburg, Germany

³Siemens Healthcare, Forchheim, Germany

Address correspondence to: Dr Nils Petri

E-mail: petri_n@ukw.de

Objective: To detect an in-stent restenosis, an invasive coronary angiography is commonly performed. Owing to the risk associated with this procedure, a non-invasive method to detect or exclude an in-stent restenosis is desirable. The purpose of this study was to evaluate the influence of cardiac motion on stent lumen visibility in a third-generation dual-source CT scanner (SOMATOM Force; Siemens Healthcare, Forchheim, Germany), employing a pulsatile heart model (CoroSim®; Mecora, Aachen, Germany).

Methods: 13 coronary stents with a diameter of 3.0 mm were implanted in plastic tubes filled with a contrast medium and then fixed onto the pulsatile phantom heart model. The scans were performed while the heart model mimicked the heartbeat. Coronary stents were scanned in an orientation parallel to the scanner z-axis. The evaluation

of the stents was performed by employing a medium sharp convolution kernel optimized for vascular imaging.

Results: The mean visible stent lumen was reduced from $65.6 \pm 5.7\%$ for the stents at rest to $60.8 \pm 4.4\%$ for the stents in motion (p -value: <0.001).

Conclusion: While the difference in lumen visibility between stents in motion and at rest was significant, the use of this third-generation dual-source CT scanner enabled a high stent lumen visibility under the influence of cardiac motion. Whether this translates into a clinical setting has to be evaluated in further patient studies.

Advances in knowledge: The employed modern CT scanner enables a high stent lumen visibility even under the influence of cardiac motion, which is important to detect or exclude an in-stent restenosis.

INTRODUCTION

Ischaemic heart disease is still a major cause of death in the world.¹ To treat a severe coronary artery disease, the implantation of a stent is commonly performed. Advances in stent technology have reduced the rate of in-stent restenosis; however, they still occur in the range of 3–20%.² Patients who underwent stent implantation and present themselves with recurring chest pain are often referred to invasive coronary angiography for the detection of an in-stent restenosis or progression of coronary heart disease. Because in many of these procedures (23%) an in-stent restenosis or a new significant coronary artery lesion is excluded, the patient underwent an unnecessary examination and was exposed to its associated risk.³ Thus, a non-invasive method is desirable.

CT angiography (CTA) may provide a feasible solution, offering direct visualization without the invasiveness of a coronary angiogram. Early CT generations did not offer

the required image quality.^{4,5} However, through increasing number of detector rows, increasing temporal resolution and improved reconstruction kernels, the image quality has been substantially improved.^{6–16}

A previous study revealed that a third-generation dual-source CT scanner (SOMATOM Force; Siemens Healthcare, Forchheim, Germany) enabled a lumen visibility of up to 80% in metal stents while the stents were at rest.¹⁷ To evaluate the influence of cardiac motion on stent lumen visualization in the same third-generation dual-source CT scanner, we employed a pulsatile heart model (CoroSim®; Mecora, Aachen, Germany) (Figures 1 and 2) to evaluate differences in visible stent lumen diameter between motion and standstill.

METHODS AND MATERIALS

Stent selection and experimental setup

13 coronary stents with differing stent structures and materials were examined in this study. The diameter of

Figure 1. A pulsatile heart model (CoroSim®; Mecora, Aachen, Germany).

each stent was 3.0 mm. All stents were implanted in polyolefin plastic tubes and then filled with a contrast medium diluted to a density of 350 HU at 120 kV (Imeron® 300; Bracco Imaging, Konstanz, Germany). The stents were expanded with the pressure defined by the manufacturers *in vitro* stent compliance table for the intended inner diameter of 3.00 mm. In most stents, it was their nominal diameter; only the Taxus Liberté® (Boston Scientific, Boston, MA), Cypher® (Cordis, Hialeah, FL) and Orsiro® (Biotronik, Berlin, Germany) stents were expanded to an inner diameter of 3.0 mm instead of their nominal diameter of 2.75 mm. Tubes not exactly fitting the stent were carefully heated to shrink the tube inner diameter until it matched the outer diameter of the stent while keeping the balloon inflated.

These plastic tubes, mimicking the coronary artery, were then attached to the phantom heart model (CoroSim®; Mecora).¹⁸ This phantom model is an anatomic heart model made out of silicone, filled with fluid and driven by a pump for the pulsatile flow. This flow causes the alternating expansion and contraction of the model, which is comparable with a beating heart in humans. The phantom heart rate is fixed to 74/min. Thus, for electrocardiogram (ECG) triggering of the CT, an ECG simulator was connected to the scanner with a corresponding frequency of 74/min.

Figure 2. A close up of the heart model with the tube containing the stent attached.

CT scan protocol

The study was performed on a third-generation dual-source CT system (SOMATOM® Force; Siemens Healthcare).

Scans were performed with the stents in an orientation parallel to the scanner z-axis and acquired in a retrospectively ECG-gated cardiac spiral (collimation 96×0.6 mm, tube voltage 120 kVp, 340 mAs reference tube current). The number of acquired slices is virtually doubled by means of a diagonal flying focal spot yielding a 192×0.6 -mm acquisition at a gantry rotation time of 250 ms. Automatic exposure control was active during the acquisition, enabling the adjustment of the tube current to the size of the phantom based on the topogram information, thus allowing clinically more realistic noise impressions of the images. ECG-based tube current modulation was not applied to enable the reconstruction of images with the same noise for all time points of the phantom motion cycle.

Data reconstruction

All CT images were reconstructed with a third-generation iterative reconstruction technique (ADMIRE; Siemens, Forchheim, Germany) using an image slice thickness of 0.6 mm, an increment of 0.3 mm and field of view of 170 mm with an image matrix of 512×512 pixels and a temporal image resolution of 66 ms. Each data set was reconstructed with a medium sharp convolution kernel optimized for vascular imaging (Bv49, strength 3; possible strength settings from 1 to 5).

The Bv49 kernel is in its intended use equivalent to a B46f coronary stent evaluation kernel on the second-generation dual-source CT scanner SOMATOM Definition Flash (Siemens Healthcare, Forchheim, Germany) with 16% sharpness increase and modulation transfer function parameter $\rho_{50} = 5.62 \text{ lp cm}^{-1}$ and $\rho_{10} = 7.55 \text{ lp cm}^{-1}$. It has to be noted that the selected kernel operates below the theoretical resolution limits of the system; however, as it is applied in a cardiac scan acquisition, which is optimized for the highest temporal resolution based on a partial scan-type reconstruction, this kernel reflects the best trade-off between the maximum applicable sharpness and the clinically acceptable image noise.

Image analysis

Axial reformations of all stents were used for evaluation using a window width of 1500 HU and a centre of 300 HU. A secondary longitudinal multiplanar reformation was generated for demonstration purposes only (Figure 3). The visible lumen diameter was assessed at a static phase at maximum expansion and at maximum motion, defined as midway between maximum expansion and contraction. Measurements were performed on three representative axial slices each, using the electronic caliper tool provided with the manufacturer three-dimensional reading software (Syngo®.via). Attenuation inside the visible stent lumen was evaluated on the same three axial slices using a region of interest measurement as large as possible not including the stent strut artefacts (Figure 4).

Statistical analysis

Normal distribution was evaluated by the Kolmogorov–Smirnov test and the Shapiro–Wilk test. Based on these tests, normal

Figure 3. A visible representation of the phases of the heartbeat of the model and its influence on the stent lumen diameter.

distribution was assumed for visible stent lumen diameter. The measurement results of these variables are provided as mean, standard deviation and range. Statistical significance was assumed at $p < 0.05$ and evaluated with the Student's t -test in the normally distributed variables. The interobserver agreement was determined with the Pearson correlation.

RESULTS

Influence of motion on visible stent lumen diameter
The mean visible stent lumen diameter in the static phase of the phantom model was $65.6 \pm 5.7\%$, ranging from 54.4% in the Kaname® (Terumo, Tokyo, Japan) to 76.7% in the Taxus Liberté®. At maximum motion, the mean visible stent lumen diameter was $60.8 \pm 4.4\%$, ranging from 51.1% in the Kaname® to 66.7% in the JOSTENT Graftmaster® (Abbott). This difference in visible stent lumen between static and motion phases proved to be highly significant with $p < 0.001$. Mean visible stent diameters in the static and motion phases are provided in Table 1. Visible stent lumen diameters in the static and motion phases of the individual stents are provided in Table 2. Interobserver agreement was calculated for a subset of the data (for 5 of the 13 stents) and was found to be very good with a Pearson correlation coefficient of 0.963 ($p = 0.001$).

Influence of motion on attenuation

The mean in-stent attenuation in the static phase was 346 ± 29 HU, ranging from 311 HU in the Cypher® to 405 HU in the

JOSTENT Graftmaster®. At maximum motion, the mean in-stent attenuation was 397 ± 66 HU, ranging from 268 HU in the Omega® (Boston Scientific, Boston, MA) to 549 HU in the Coroflex Blue® (Braun, Kronberg im Taunus, Germany). This difference with more realistic attenuation values in the static phase, *i.e.* attenuation closer to the actual contrast density of 350 HU, proved to be significant with $p = 0.004$.

Attenuation in the unstented tube was 385 HU in the static phase and 380 HU at maximum motion. With $p = 0.48$, this difference between standstill and motion proved not to be statistically significant.

DISCUSSION

This *in vitro* study analyzed the in-stent lumen visibility of 13 commonly used stents evaluated by the utilization of a third-generation dual-source CT scanner (SOMATOM Force; Siemens Healthcare, Forchheim, Germany), while a pulsatile heart model (CoroSim®; Mecora) simulated the heartbeat and thus the associated motion of the stent. The third-generation dual-source CT scanner offers an improved temporal resolution, which should reduce motion and blurring artefacts. Blooming artefacts caused by X-ray attenuation of the stent impair the lumen visibility. In addition to this, motion artefacts cause further reduction in lumen visibility. In previous studies, especially in patients with heart rates >70 beats min^{-1} , artefacts were

Figure 4. A measurement example of in-stent lumen visibility and attenuation. Max., maximum; Min., minimum.

Table 1. Mean visible lumen diameters and in-stent attenuation in the static and motion phases

Parameter	Static	Motion	<i>p</i> -value
Visible diameter (%)	65.6 ± 5.7 (54.4–76.7)	60.8 ± 4.4 (51.1–66.7)	<0.001
Attenuation (HU)	346 ± 29 (311–405)	397 ± 66 (268–549)	0.004
Noise	27.0 (26.5–28.0)	29.0 (27.0–36.5)	0.012

Values of visible diameter and attenuation are provided as mean, standard deviation and range.

marked.⁵ Moreover, owing to these artefacts, impairment of image quality is significantly correlated with increasing heart rate.¹⁹ To counter this effect, beta-blockers are commonly used.^{6,20} However, in previous studies, even with the application of oral or i.v. beta-blockers, target heart rates of <65 beats per minute could not be reached in all patients.²¹ Motion artefacts at higher heart rates are especially hindering the evaluation of the mid-right coronary artery in clinical practice, where the translational movement during contraction and relaxation of the heart is of particular extent.²² Therefore, a preferably high temporal resolution is desirable in patient studies, allowing for the evaluation of coronary artery segments even in the presence of cardiac movement.

We found the mean visible stent lumen, *i.e.* the average degree of narrowing of the known true lumen diameter in percentage, in the static phase of the phantom model to be 65.6 ± 5.7%. At maximum motion, the mean visible stent lumen diameter was reduced to 60.8 ± 4.4%. The difference in visible stent lumen between static and motion phases proved to be significant; however, the difference between the mean visible stent lumen in the static phase (65.6 ± 5.7%) and at maximum motion (60.8 ± 4.4%) was less than the range between the different stents (54.4–76.7%), showing that the effect of motion is smaller than the effect of the stent design.

In the present study, we chose a tube voltage of 120 kVp. In a previous study, we could show that the highest values of stent

lumen visibility can be observed at 120 kVp in the employed third-generation dual-source CT. When choosing lower tube voltages, stent lumen visibility is slightly reduced.²³

We have measured the image noise; however, we think that the coronary flow of the heart model contributes to the artefacts, thus rendering this parameter non-valid.

The current third-generation dual-source CT scanner provided a high stent lumen visibility of 60.8 ± 4.4% at motion. As in clinical routine stenoses are usually only considered to be of haemodynamic significance when reaching 50% or more, coronary CTA may be a valid method for detecting in-stent restenosis even during cardiac motion phases. In general, improving the accuracy of coronary CTA might help reduce the number of solely diagnostic coronary angiographies, which are invasive and have a higher radiation dose. Whether this translates into a clinical setting has to be evaluated in further patient studies.

Limitations

In our setup, only cardiac motion was simulated; respiratory movements were not simulated. Motion of the phantom model as well as of the attached stent is only an approximation to conditions *in vivo*. Also, artefacts caused by arrhythmic heartbeats were not analyzed in this study. Furthermore, the effect of the surrounding torso on stent visualization *in vivo* has to be considered. In our

Table 2. Visible stent lumen diameters of individual stents at motion and standstill

Manufacturer	Stent	Diameter (mm)	Visible diameter (motion) (%)	Visible diameter (static) (%)
Boston Scientific (Boston, MA)	Taxus Liberté®	3.0	62.2	67.8
Cordis (Hialeah, FL)	Cypher®	3.0	64.4	76.7
Abbot (Chigaco, IL)	JOSTENT Graftmaster®	3.0	66.7	67.8
Orbus Neich (Hong Kong, Japan)	Genous	3.0	63.3	70.0
Terumo	Nobori	3.0	63.3	70.0
Terumo (Tokyo, Japan)	Kaname®	3.0	51.1	54.4
Braun (Kronberg im Taunus, Germany)	Coroflex Blue®	3.0	54.4	66.7
Biotronik	PRO-kinetic energy	3.0	58.9	60.0
Abbott (Chigaco, IL)	Multilink vision	3.0	62.2	65.6
Medtronic (Dublin, Ireland)	Resolute RX	3.0	60.0	64.4
Boston Scientific (Boston, MA)	Omega®	3.0	60.0	61.1
Medtronic (Dublin, Ireland)	Integrity RX	3.0	65.6	68.9
Biotronik (Berlin, Germany)	Orsiro®	3.0	57.8	60.0

setup, the stent and the tubes were filled with a contrast medium diluted to 350 HU at 120 kV, but there was no material surrounding the tubes. In addition, as tubes were not always available in sizes exactly matching the intended stent diameter, some of them were carefully heated. To minimize a possible effect of the shrinking process on stent lumen diameter, this procedure was performed while keeping the balloon inflated. Although no visible differences in stent lumen diameter were observed, a possible effect on the stent lumen diameter due to heating of the tubes cannot be totally excluded. A fixed window setting with a width of 1500 HU and a centre of 300 HU was chosen and a single observer performed the caliper measurements. Although a subset of the data was analyzed by a second observer, measurement-related errors and the effect of different window settings on blooming and visible stent lumen diameter have to be taken into account.

CONCLUSION

This study evaluated the influence of cardiac motion on stent visualization in a third-generation dual-source CT scanner employing a pulsatile heart model. While the difference in lumen visibility between stents in motion and at rest is significant, the use of this third-generation dual-source CT scanner enables a high stent lumen visibility under the influence of cardiac motion. Whether this translates into clinical practice and helps to reduce the number of solely diagnostic coronary angiographies has to be evaluated in further patient studies.

CONFLICTS OF INTEREST

Allmendinger and Flohr are employees of Siemens Healthcare, Forchheim, Germany.

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